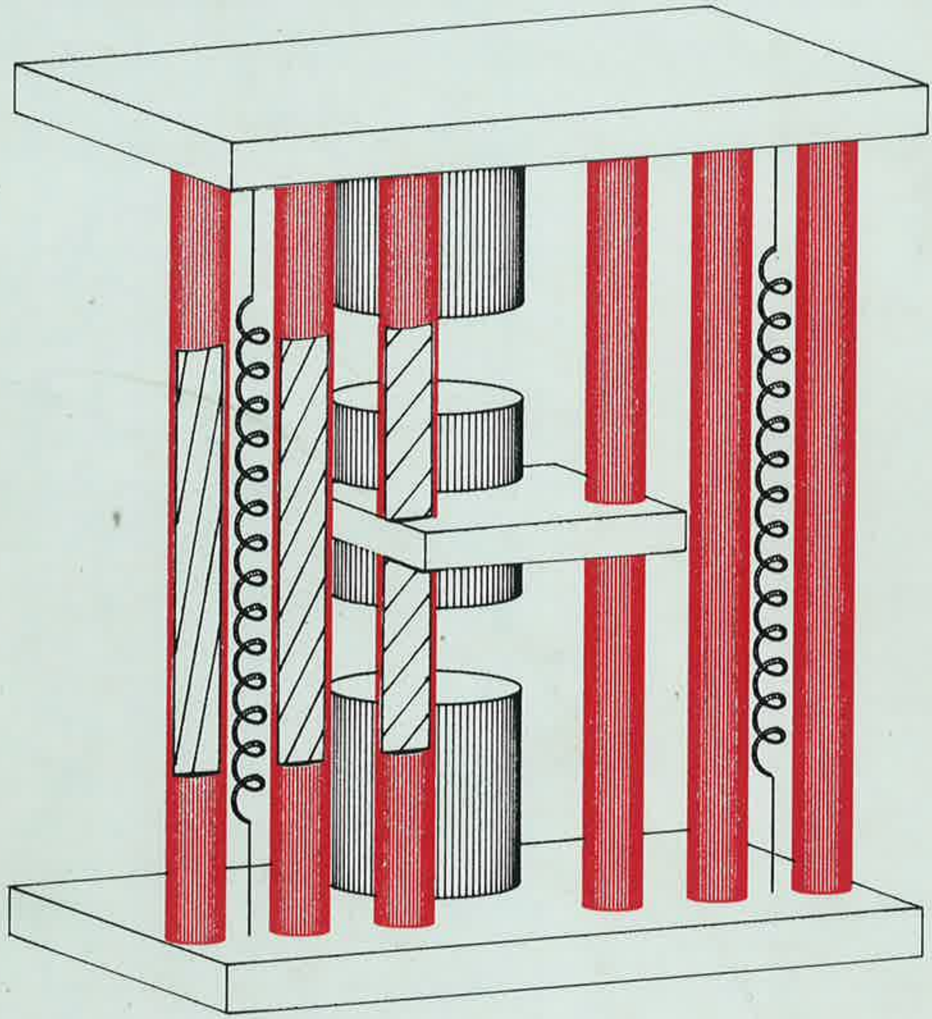
Functions of the lumbar back muscles



P.Vink

Functions of the lumbar back muscles

Proefschrift

ter verkrijging van de graad van Doctor
aan de Rijksuniversiteit te Leiden,
op gezag van de Rector Magnificus
Dr. J.J.M. Beenakker,
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Peter Vink

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Stellingen
behorende bij het proefschrift

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Andersson et al. stellen dat er geen significante verschillen zijn gevonden tussen de drie onderdelen van de m.erector spinae. Wanneer de door Andersson et al. gevonden EMG waarden worden uitgezet als functie van de inclinatiehoek bij het zitten, zijn de verschillen wel significant.

Andersson et al., 1974, Scand J Rehab Med Suppl 3, 91-108

Het verschil in vezeltype verklaart in onvoldoende mate het verschil tussen recrutering van de drie kolommen van de intrinsieke lumbale rugspieren.

Dit proefschrift

De maximale kracht, die nodig is om het dreigende voorwaarts en zijwaarts vallen van de romp tijdens lopen te voorkomen, wordt passief geleverd door rek van de intrinsieke lumbale rugspieren en/of het omliggende weefsel.

Dit proefschrift

De uitwendige belasting van de wervelkolom is niet de oorzaak van rugklachten, maar het is de belasting in kombinatie met de manier waarop het individuele lichaam de belasting opvangt.

Dit proefschrift

Volgens Bogduk (1980) is de lumbale intermusculaire aponeurose gelocaliseerd tussen de m.longissimus thoracis en de m.iliocostalis lumborum. Dit geldt maar voor een zeer beperkt gebied.

Bogduk, 1980, J Anat 131, 525-540

In een zeer beperkt aantal situaties zal een beenlengteverschil tot een bekkenscheefstand leiden.

Dit proefschrift

Een verhoogde stenentas en speciekuip in combinatie met een metselnivo boven kniehoogte (te realiseren door een hefsteiger met ingebouwde verhoging voor stenentas en speciekuip) verlaagt de energetische belasting voor het merendeel van de metselaars tot een aanvaardbaar nivo.

Arbouwpublicatie, Amsterdam juli 1988

Clinici zien alleen die mensen met een beenlengteverschil, die klachten of aandoeningen hebben. De beeldvorming van clinici zal dus bijgesteld moeten worden.

De theorie van Gal'perin geeft waardevolle aanwijzingen om het vak anatomie zo te structureren dat het leerproces van de student efficienter en effectiever verloopt.

De vakdidactiek van de anatomie dient aangepast te worden aan het toegenomen functioneel-anatomisch inzicht.

Arbeidsomstandigheden kunnen in biomechanische zin pas goed wetenschappelijk gemeten worden wanneer dit weinig relevant is.

Musisch-agogische kennis behoort in alle instituten die opleiden en voorlichten toegepast te worden.

Met 'volgende week zondag' wordt meestal hetzelfde bedoeld als 'aanstaande zondag'. Een van de uitdrukkingen is dan waardeloos.

To my mother,
my wife Marianne,
Efraim and
Ruben.

Cover figure
Important functions and activities of the intrinsic lumbar back muscles (=ILBM) are presented in a model:

The six columns in the model represent longitudinally arranged subdivisions which could be distinguished in the intrinsic lumbar back muscles, three on each side of the spine. These have approximately equal momentarms in the sagittal plane and different momentarms in the frontal plane.

Shading in the columns represents the difference in relative contribution of each column to an extension force, which is probably due to differences in muscle architecture.

Springs between the columns represent elastic properties in series or parallel, which are apparent in the intrinsic lumbar back muscles and surrounding tissue.

Three parallel planes: The lateral columns are represented in this model as units running between thorax (upper plane) and pelvis (lower plane) because these muscles are recruited as a whole. The medial columns show a different behaviour for at least two distinguishable lumbar levels within the column and may be able to move the lumbar vertebrae seperately.

Contents

1. Introduction

1.1 Back muscles and back pain prevention

An impressive amount of research in the field of back pain, is focussed on curative, biomechanical, ergonomical, aetiological or epidemiological aspects. Yet, it is still impossible to reduce costs concerning low back pain (now about 800.000 guilders per work hour in The Netherlands).

In spite of the impressive amount of research, I suppose that the precise cause of back pain will remain unknown for several decennia. Not only because the back is anatomically and functionally one of the most complex parts of the human body, but above all because back pain is a symptom resulting from many different pathologies while each of them may have a multifactorial aetiology. Meanwhile an attempt has been made to prevent back pain by diminishing heavy loadings.

However, low back pain occurs both in physically 'heavy' and 'light' occupations. Only marginal differences in prevalence between nurses and teachers and between manual and office workers were found (Dales et al., 1986). Thus, not loading itself is a major problem, but the way in which each individual human body withstands this loading. This also explains why some never have back pain in spite of their 'heavy' work. The problem is still more complicated because there is no simple relationship between subjective complaints of low back pain and objective clinical evidence of structural damage.

The only active way to withstand a particular loading without damage is by muscle recruitment. Therefore knowledge of the functions of the lumbar back muscles is needed in the field of back pain prevention. This is stressed by the finding that in heavy occupations only about 20% of the maximum voluntary contraction (=MVC) of the back muscles in static and continuous work can be sustained (Legg & Pateman, 1983). This also stresses an important boundary condition to endure loadings without damage: excessive loadings should be avoided. Therefore the problem how to prevent an occupational, structural and/or functional damage of the low back should be tackled from at least two different sides, while it must be realized that both approaches are interrelated. One approach follows the line of establishing limits to the magnitudes of loads imposed on the back during work, the other is directed to the optimalisation of handling such loads by training muscle control as well as (re)designing the work environment. This thesis is focussed methods for the optimalisation approach.

Indeed, whatever the cause of low back pain may be, back pain can be prevented by teaching handling methods in combination with environmental adaptation. Subjects must be trained in the optimal use of their muscles, and the environment must be (re)designed to enable this optimal use. However, the optimal use of muscles (optimal is without damage), is unknown. Thus, there is a strong need for knowledge of the optimal recruitment of the back muscles in given postures as well as movements, because these are the active elements that control the loading of the lumbar back. Back pain free subjects who perform hazardous activities over years are of special interest. Knowledge of their pattern of muscle recruitment may provide indications for an optimal use.

Before studying the optimal activity in a complex occupation, a method should be developed and it should be tested whether this method can be used in a number of dynamic and static activities. The development of the method as well as the tests are described in this thesis.

1.2 Purpose of the investigations

The aim of the research presented in this thesis was primarily to study the way lower back muscles can stabilise the spine under various loading conditions. To this aim the recruitment pattern of lumbar back muscles was recorded electromyographically in static and dynamic situations. Parts of these situations may cause low back pain in the long run, such as exerting maximal muscle forces or standing and walking with artificial leg length discrepancies up to 40 mm.

The activity of different parts of the lumbar back muscles was recorded separately to establish the relative contribution of these parts under various loading conditions. A pilot study already revealed significant differences in recruitment between two parts of the lumbar back muscles (Dofferhoff &

Vink, 1985). However, whether such a distinction could be refined further was unknown. Hence the anatomy of the lower back muscles (see chapter 2) as well as the selectivity of the available methods and equipment (see chapter 3) were studied firstly. The activity of three columns of the ILBM (see box I) was then studied at different force levels from zero up to maximum.

BOX I: THE MUSCLES STUDIED
The largest muscle group of the back, the intrinsic lumbar back muscles (=ILBM), was studied, because these were able to exert the largest force. Intrinsic means that these muscles essentially belong to the back: ontogenetically these muscles develop from the dorsal region of the myotome and they are innervated by the dorsal rami of the spinal nerves. Three different columns of the ILBM could be studied: a medial (containing the multifidus, the spinalis and partly the longissimus muscle), an intermediate (longissimus muscle) and a lateral column (iliocostalis muscle).

ILBM activity was also studied during walking. In an attempt to understand mechanically the pattern of muscle activity during walking, kinematic information concerning segments directly connected to the lumbar back muscles had to be added. This required the development of a technique for measuring pelvic rotations during walking (Vink, 1986) and synchronisation of the digitised signals (EMG, pelvic rotations and stride times).

After recording activities of the lumbar back muscles during normal walking, mechanisms stabilising the spine were studied during standing and walking with an artificial leg length difference. This approach was chosen because it is often claimed that leg length differences cause a pelvic tilt during many postures and movements which may result in low back pain (f.i. Helliwell, 1985). Whether this is true will be discussed in the sixth and seventh chapter.

1.3 Relationship with other projects of the research group

The experiments were chosen in alignment with developments in the laboratory of the interdepartmental research group of kinesiology. This laboratory was equipped for electromyographic studies as well as for human walking on a treadmill. Previous studies concern activity of the lower leg muscles during walking (Ambagtsheer, 1978), the quantification of cross-talk between signals of different muscles (Morrenhof and Abbink, 1985) and the recording of gait parameters on a treadmill made of conducting rubber (Kauer et al., 1985).

1.4 The pilot study

Dofferhoff and Vink (1985) used the treadmill in combination with EMG recordings to study the functions of the mm. multifidi and the mm. iliocostales lumborum. Bipolar surface electrodes were applied bilaterally on these muscles. Recordings were made at two different running speeds of the treadmill and under varying loading conditions of the spine.

Both the multifidi and the iliocostales showed an increase in duration of activity relative to unloaded walking when the subjects carried a load of 5 kg in front of the trunk and a decrease of activity time with a load of 5 kg on the back. A load on the lateral side of the body resulted in a decreased duration of multifidus activity on the homolateral side. In three subjects the same pattern was found for the iliocostales, but in the other three subjects there was no activity at all. At the contralateral side, the duration of iliocostalis activity was increased, while the multifidus muscles demonstrated no increase. Thus, the iliocostalis muscle seems to have a more important function in counteracting a lateral load on the torso then the multifidus.

1.5 Experiments of this thesis

In order to study how the spine is stabilised by the ILBM, knowledge of the activities of the major parts of these muscles was needed. Therefore activities of three columns of the ILBM at different lumbar levels were recorded with 12 bipolar surface electrodes. Surface electrodes were used because they have a large pick up area and because they are not invasive. However, surface electrodes also pick up signals from muscles for which the electrodes are not meant (=cross-talk). The extent of this problem is quantified in chapter three, where the cross-correlation coefficient function (=CCCF) is used as a measure of the occurring cross-talk.

In chapter four the contribution of the three columns was studied at different force levels up to maximum in a static situation. As extension is the major effect of the ILBM, EMG was recorded during extension of the lumbar back against a mechanical resistance in the upright position. The upright position was chosen to exclude the gravitational effect as much as possible. To this aim we developed an experimental set-up in which the extension force was measured externally with a spring balance, which had been fixed at a girdle around chest and arms, while the subjects stood with the pelvis and thighs against a board to push off. The rectified and averaged EMG (abbreviated to RA-EMG) was calculated for each of the three columns to see differences between the individual parts.

In chapter five the activity of the ILBM was studied under a dynamic condition: walking. In dynamic situations recording of the amplitude of the EMG is insufficient, because muscle work is also influenced by the period of activity. During walking this

BOX II: RA-EMG
RA-EMG is comparable to other terms used in the literature (IEMG = integrated EMG, SRE-EMG = smoothed rectified EMG, RMS = root mean square and FRA-EMG = fullwave rectified and averaged EMG). A new term RA-EMG is introduced because the terms in the literature are used for other techniques (f.i. electronic processing). In our case firstly the absolute (rectifying) of the digitised raw EMG signal is computed and then the average over a period of time is calculated (averaged). In future it is essential to use the terms which are to be formulated by the International Society of Electrophysiology and Kinesiology in 1990 at Boston University.

period of activity is determined by for instance the length of a stride. Therefore EMG was related to stride times and pelvic rotations. Registration of stride times was realized on a treadmill made of conducting rubber while the subjects wore shoes with contacts indicating heel strike and toe off for the left as well as for the right foot. The rotations of the pelvis in the frontal and sagittal plane were recorded simultaneously with a 'pelvis girdle', which was firmly strapped onto the pelvis (Vink, 1986; Van Leeuwen et al., 1988). The 'pelvis girdle' was connected to two potentiometers, which recorded deviations from a horizontal position. The activity pattern of each

EMG lead and the pelvic rotation was afterwards calculated on a PDP 11/70 computer for different parts of the stride in order to study the stabilising functions of the ILBM.

In chapter six a 3-D computer model was developed to study the role of forces generated by passive lengthening of the three columns of the ILBM. Data concerning the rotations of the pelvis and the thorax were the input in the model, lengths of the columns the output. This model offers an explanation for the observation that EMG is not always synchronous with muscular force deliverance.

In chapter seven the relationship between pelvic tilt and ILBM activity was studied during standing. Recordings were made, while the subjects stood upright with extended knees. An artificial leg length discrepancy was created by putting different boards of 5.0 mm under the right foot. Pelvic angles were recorded with the pelvis registration system and recorded simultaneously with the EMG. The effect of an artificial leg length discrepancy on ILBM activity was then described.

In chapter eight the influence of an artificial leg length discrepancy on ILBM activity was studied during walking. To this aim raised shoes (10, 20, 30 and 40 mm) were made with a sole which allowed the foot to roll off and with contacts indicating heel strike and toe off.

In both latter chapters predictions based on the results from chapter three to six were verified and data were collected concerning the possible influence of a leg length inequality on low back pain.

1.6 Future research

The method in this thesis is developed to be used in prevention of back pain producing results at short notice. In continuation of this thesis a systematic approach of practice oriented research is suggested.

I The following stages have to be followed for several hazardous activities:

1. A hazardous activity (or a series of hazardous activities performed professionally or in sports) should be selected, based on epidemiological data, on interviews or questionnaires.

2. Observations or task analysis of the hazardous activity should be made. Three types of subjects should be compared: back pain free subjects performing hazardous activities over years, back pain patients performing hazardous activities over years and back pain free subjects which never have performed the hazardous activity. In case obvious differences are found, preventive actions may start.

3. Laboratory research is needed if the observation or the job analysis do not show significant differences between the three groups or if the data are essential for general directives (see II). The laboratory research includes EMG-analysis of the three groups of subjects with additional information such as kinematic data, force recordings or estimations by biomechanical models as mentioned in this thesis.

4. Training and environmental adaptations should be undertaken based on the recruitment of an optimal movement (derived from 3) or on differences in handling methods (derived from 2).

5. The effect of the prevention activities should be evaluated by for instance epidemiological methods.

II. When a number of hazardous activities are investigated, a general theory should be deduced from several cases and tested in the succeeding case. The theory concerns the optimal recruitment of the lumbar back muscles related to postures and movements of anatomical structures. This theory directs the training and environmental adaptations, which should be known by for instance occupational health services and ergonomists and in a simple form even by the subjects themselves.

8

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2. Morphology of the intrinsic lumbar back muscles.

2.1 Introduction

In this chapter the anatomy of the intrinsic lumbar back muscles (=ILBM) is described. Knowledge of the anatomy of the ILBM is essential to attribute the recorded activity to a particular anatomical subdivision. If the activity is attributed to an anatomical subdivision, the effect of a contraction can be predicted and the contribution of a particular part in stabilising the spine can be identified.

The description of the ILBM in standard textbooks of anatomy (e.g. Braus, 1954; Warwick & Williams, 1973) is based on earlier studies such as Eisler (1912) and Winckler (1948). However, some renewed descriptions have been presented recently. These are given also attention in this chapter.

2.2 The lumbar back muscles

2.2.1 The thoracolumbar fascia

The thoracolumbar fascia covers the lumbar back muscles. It arises from the dorsal side of the iliac crest, the sacral and lumbar spinous processes and the intervening supraspinous ligaments. Superiorly the fascia shades off partly into muscles (this part is in fact an aponeurosis) or attaches to the thoracic spinous processes and the angles of the ribs.

The fascia covers two large muscles (nomina anatomica, 1966): laterally the erector spinae muscle and medially the multifidus muscle.

2.2.2 The erector spinae muscle

The erector spinae muscle consists of two separate muscles, partly separated by a septum: the longissimus thoracis and the iliocostalis lumborum.

2.2.2.1 The intermuscular septum in the erector spinae muscle.

Bogduk (1980) claims that both parts of the erector spinae muscle are separated by a septum, which is named the lumbar intermuscular aponeurosis by Bogduk. The longissimus lies medially to the lumbar intermuscular aponeurosis and the iliocostalis laterally. However, this septum was already mentioned by for instance Langenberg (1970) and Jonsson (1970).

According to Bogduk (1980) the lumbar intermuscular aponeurosis arises from the ilium and runs essentially parasagittal in cranial direction. The dorsal part is continuous with the thoracolumbar fascia. The ventral edge reaches almost to the dorsal surfaces of the lumbar transverse processes, but is separated from them by a fat-filled space. However, this continuity ventrally is not shown in the study of Jonsson (1970) (see fig 2.2). The connective and muscle tissue becomes very irregular ventrally and whether an aponeurosis can be distinguished ventrally seems doubtful. Moreover, the study of Jonsson (1970) shows only at levels L1, L2 and L3 an aponeurosis located between the muscles. Accordingly, Bustami (1986) showed that at lower lumbar levels the aponeurosis lies laterally to the whole erector spinae.

Thus, only at cranial lumbar levels in the dorsal part of the erector spinae muscle the lumbar intermuscular aponeurosis can be seen as one clearly separated septum (see fig. 2.2).

2.2.2.2 The longissimus muscle

The longissimus muscle should be divided into two parts according to Bustami (1986): a caudal and cranial part respectively (see fig. 2.1).

The caudal part of the longissimus muscle is the deeper part of the erector spinae muscle in the lumbar region. It arises from the ilium and from the anterior and medial sides of the lumbar intermuscular aponeurosis. Standard textbooks (e.g. Romanes, 1972; Warwick & Williams, 1975) show or report wrongly an origin from the spinal or transverse processes of the lumbar vertebrae of this part of the muscle. The muscle runs in a medial-cranial direction and inserts into the tips and accessory process of the trans-

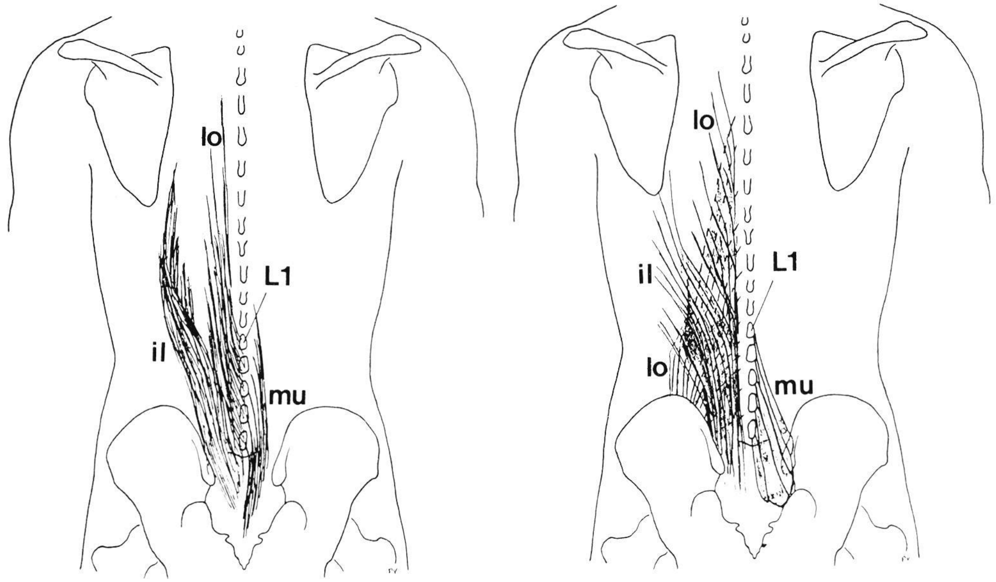

Fig 2.1. Traditional description of the intrinsic lumbar back muscles (left) and the description according to recent studies (Bogduk, 1980; Bustami, 1986; Macintosh et al., 1986)(right). MU= m.multifidus, LO= m.longissimus, IL= m. iliocostalis.

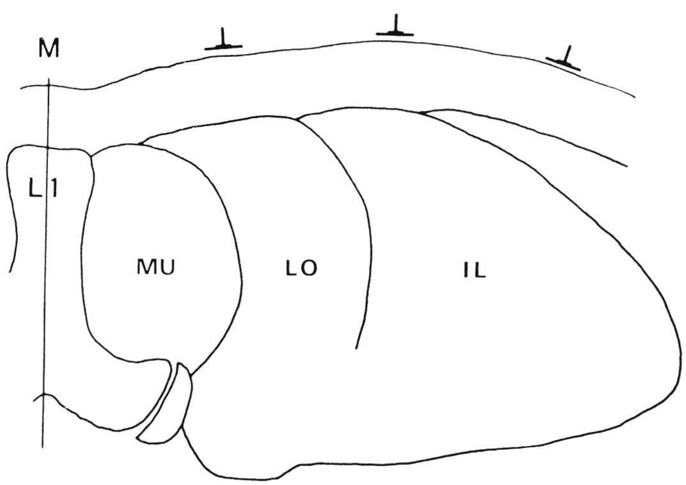

Fig. 2.2. Three columns of the intrinsic lumbar back muscles (ILBM) in a transverse section through the spinous processus of L1 according to Jonsson (1970). M= median plane, MU= m.multifidus, LO= m.longissimus, IL= m.iliocostalis

verse processes of lumbar vertebrae L1 to L5 and thoracic vertebrae Th4 to Th12 (Bustami, 1986). Bogduk (1980) and Langenberg (1970) describe this muscle part similarly, but consider the fibers inserting into the lumbar accessory process as the deeper part of the iliocostalis muscle. However, the accessory process is a part of the transverse process medially to the attachments of the longissimus muscle, while ilio<u>costalis</u> suggests a lateral insertion. Therefore this is considered as a longissimus muscle like Bustami (1986) does.

The cranial part of the longissimus muscle lies superficially and partly overlaps the caudal part. It arises from the anterior side of the thoracolumbar fascia and sometimes from the transverse processes of the vertebrae Th6-Th12 (Langenberg, 1970; Bogduk, 1980; Bustami, 1986). The eight slips run latero-cranially and insert into the third to tenth ribs (Bustami, 1986). This cranial part should be distinguished from the caudal part, because the laminae differ in size and terminations and thus in function. A contraction of the cranial part would have an effect on the ribs, while the caudal part would act on the vertebrae.

2.2.2.3 The iliocostalis lumborum muscle.

The iliocostalis lumborum muscle arises from the iliac crest and the sacrum by way of the lumbar intermuscular aponeurosis and the thoracolumbar fascia. It is attached to the posterior and lateral margins of the aponeurosis and inserts by slender flattened tendons into the angles of the inferior ten ribs (Bustami, 1986). The fibers of the iliocostalis muscle extend from caudomedial to laterocranial and cross the fibers of the caudal part of the longissimus muscle posteriorly (see fig. 2.1), because the longissimus fibers extend deep from caudolateral to mediocranial (Bustami, 1986). Although inserting at different vertebral levels, there is no line of demarcation between lumbar and thoracic fibers (Bogduk, 1980).

2.2.3 The multifidus muscle

The multifidus muscle is situated medially to the erector spinae muscle and is the largest muscle that spans the lumbosacral joint. It consists of five bundles, each extending from the sacrum to a lumbar spinous processus. The bundles which arise medially from the sacrum insert into to the lower lumbar vertebrae. The bundles which arise more laterally and superiorly from the sacrum , the iliac tuberosity and the mammillary process (a small tubercle between the transverse and the spinous process) of lowest three lumbar vertebrae (Langenberg, 1970; Macintosh et al., 1986), insert into higher lumbar vertebrae (see fig. 2.1). The distinction in five bundles as well as the origin from the iliac tuberosity is absent indeed in standard textbooks (e.g. Romanes, 1972; Warwick & Williams, 1975).

2.2.4 The weight of the ILBM at different levels

The ILBM are principally composed of the multifidus, longissimus and iliocostalis muscle in the lumbar region. The total weight of these muscles in the lumbar region is similar at each vertebral level (Etemadi, 1974), whereas the lateral border of the ILBM runs more medially towards lower lumbar levels (Jonsson, 1970). This is possible because of the increase in muscle mass anteriorly.

The weight of the iliocostalis relative to the other parts is largest at all lumbar levels. The weight of the multifidus and iliocostalis increases towards lower lumbar levels, and the weight of the longissimus decreases caudally (Etemadi, 1974).

2.2.5 Other back muscles

In the upper lumbar region the lateral part of the ILBM is covered by the latissimus dorsi and the serratus posterior inferior muscles.

Medially and under the thoracolumbar fascia the spinalis muscle is situated. This muscle is only partly situated in the superior part of the lumbar region, medially to the longissimus muscle. It arises from the spinous processes of the upper lumbar vertebrae and ascends to the thoracic spinous processes.

Deeper to the multifidus muscle are the rotatores muscles. These are small muscles connecting the transverse processus of one vertebrae to the lamina of the cranially located vertebra. These muscles are only very rarely found in the lumbar or sacral region. They are only common in the thoracic region.

Between the spines of contiguous vertebra the interspinalis muscles extend. These are small intersegmental muscles which connect two adjacent spinous processes like the intertransversarii which connect two adjacent transverse processes.

The quadratus lumborum muscle is situated latero-anterior to the transverse processus. It arises from the iliac crest and by aponeurotic fibers from the iliolumbar ligaments, which connect the ilium and the fifth lumbar vertebrae. It is attached above to the medial half of the lower border of the last rib and by four small tendons to the apices of the transverse processes of the first four lumbar vertebrae.

The psoas major muscle arises from the transverse processes, from the bodies of all the lumbar vertebrae and from the intervertebral discs. The muscle descends inside the pelvis and is attached to the lesser trochanter of the femur.

2.3 Classification of the lumbar back muscles

The back muscles in the lumbar region are classified in this thesis in intrinsic lumbar back muscles (=ILBM) and extrinsic lumbar back muscles (m.quadratus lumborum and m.psoas major). The ILBM are those back muscles which essentially belong to the back: ontogenetically these muscles develop from the dorsal region of the myotome and these muscles are innervated by the dorsal rami of the spinal nerves. These ILBM can be subdivided into the large rotator and extensor masses (m.spinalis, m.multifidus, m.longissimus and m.iliocostalis) and the short intersegmental muscles (mm. rotatores, mm.interspinales and mm.intertransversarii).

This classification differs from the traditional one (e.g. Romanes, 1972; Warwick & Williams, 1973), because these describe the erector spinae actually as one muscle. However at least, two separate muscles should be discerned: the longissimus and the iliocostalis muscle. Not only the attachments and the musculotendinous fibers can be distinguished (Bogduk, 1980; Bustami, 1986; Jonsson, 1970; Langenberg, 1970) but also the direction of the fibers is essentially different, which is functionally of significance. Neither has the nomenclature of Jonsson (1970) and Andersson et al. (1974) been followed, because it is in contradiction with the nomina anatomica (1966), which describes the m.erector spinae and the m.multifidus as two separate muscles. On the contrary, Jonsson (1970) and Andersson et al. (1974) define the multifidus muscle as the medial part of the erector spinae muscle.

2.4 The functions of the lumbar back muscles.

An isolated contraction of a muscle produces a certain motion in a joint. In this thesis this movement is defined as the mechanical effect of a muscle.

Theoretically, the mechanical effect of parts of the ILBM can be predicted from its line of action with respect to the rotation axis of a particular joint. Assuming that the axes lie in the intervertebral disc, some muscles may produce an extension (m.longissimus, m.iliocostalis, m.multifidus, m.spinalis and m. interspinalis), some a lateroflexion (m.longissimus, m.iliocostalis, m.multifidus, m.intertransversarius, m.quadratus lumborum, m.psoas maior) and some a rotation around a longitudinal axis (m.multifidus, m.rotatores). However, apart from the problem that the precise position of the axes of rotation is unknown (and thus the mechanical effect), the mechanical effect of a contracting muscle, which spans more vertebrae, is also unclear. Moreover, it is unknown which parts of the ILBM contract in the course of motion patterns.

Thus, the functions of the individual parts of the ILBM are largely unknown.

2.5 Parts of the ILBM studied in this thesis

This thesis is principally concerned with the three larger parts of the ILBM: the multifidus, the longissimus and the iliocostalis lumborum respectively. Because of the complex anatomy, for instance the crossing fibers of the m. longissimus and the m. iliocostalis, and because of the large interindividual differences, it is impossible to identify exactly the anatomical subdivisions of the ILBM recorded by the surface electrodes in vivo. Therefore the ILBM will be divided into three columns: a medial, an intermediate and a lateral column, which only grossly correspond to the multifidus, the longissimus and the iliocostalis muscles.

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3. Specificity of surface-EMG on the intrinsic lumbar back muscles

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3.1 Introduction.

The intrinsic lumbar back muscles (ILBM) have been described as functionally uniform (e.g. Battye & Joseph, 1966; Floyd & Silver, 1955; Marras et al., 1984). However, other studies show that a more refined distinction is necessary (e.g. Andersson et al., 1974; Dofferhoff & Vink ,1984; Vink et al., 1987). According to Jonsson (1970) a distinction between more than the three larger components of the ILBM -the multifidus, longissimus and the iliocostalis lumborum muscle- is hardly feasible due to the lack of accuracy in placing wire electrodes in the correct position. Although Jonsson & Reichman (1970) used various techniques to optimize electrode location, in only 75 % of their experiments a correct positioning of the wire electrodes was established.

It was shown in an EMG study of the ILBM in sitting postures (Andersson et al., 1974) that wire electrodes are more sensitive to a faulty electrode location than surface electrodes. However, surface electrodes are less selective than wire electrodes. Activity recorded with wire electrodes is only representative for a very small part of a muscle (Parker & Scott, 1973). The reproducebility and the large pick-up area, in addition to its non-invasive character, led us to the choice of surface electrodes for the EMG-records of the three aforementioned columns of the ILBM.

Unfortunately, surface EMG has a disadvantage: cross-talk (Rozendal & Meijer, 1982). The definition of cross-talk is the superposition of myoelectric activity recorded from muscles (or parts of muscles) other than for which the electrodes are meant.

Morrenhof & Abbink (1985) used the cross-correlation coefficient function (CCCF) as a quantification of the cross-talk between two electrodes. The CCCF is the changing correlation between two EMG signals recorded in the time domain with and without a variable time lag between the signals.

Using this technique it is assumed that the sources of the signals are not correlated. Furthermore, CCCF values will be increased, not only by cross-talk of striated muscles, but also by activity of the cardiac muscle (ECG) and a power-line-induced-a.c.-component (both superimposed on the two EMG signals, which are being compared).

In this paper, the specificity of surface EMG for different columns of the ILBM is assessed by calculating CCCF values. The aim of this study is to ascertain whether our experimental set-up could be used to assess the function of three columns of the ILBM and to evaluate what is the optimal electrode spacing in order to achieve this aim.

3.2 Materials and methods.

Surface EMG of the ILBM has been recorded in 12 healthy subjects (6 female, 6 male, mean age 25.5 years) in 4 different experimental situations. In each experimental situation 8 out of 12 subjects participated.

Home-made gold-plated, bipolar electrodes with a diameter of 7.5 mm and a fixed bipolar distance from center to center of 21.5 mm were used. Two electrodes were applied 30 mm and 60 mm lateral to the palpable parts of the first lumbar spinal processes (L1). A lateral electrode was placed 10 mm medial to the lateral border of the ILBM (80-90 mm lateral to the spinal processes). At level L3 two electrodes were placed at 30 mm and 60 mm and at level L5 one electrode was placed 30 mm lateral to the spinal processes (see fig. 3.1). The strongly curved skin surface in the L5 area required two separate poles of the electrodes (Hellige silver disc electrodes: type no 21717802; internal diameter 6 mm; interelectrode distance 21 mm).

The 12 signals of the ILBM, after preamplifying with differential amplifiers with a fixed gain of 100 (input impedance 100 MOhm, cmmr > 20.000, frequency response being flat between 12 Hz and 16 kHz) were led by a shielded cable to the main amplifier with variable gain up to 1000. For each of the 12 channels, the same amplification was used, so that the largest peaks were amplified to the maximum input level (+/- 1 V) of the tape-recorder (Racal Store 14D).

After low-pass filtering (1000 Hz, 24 dB/oct), the recordings

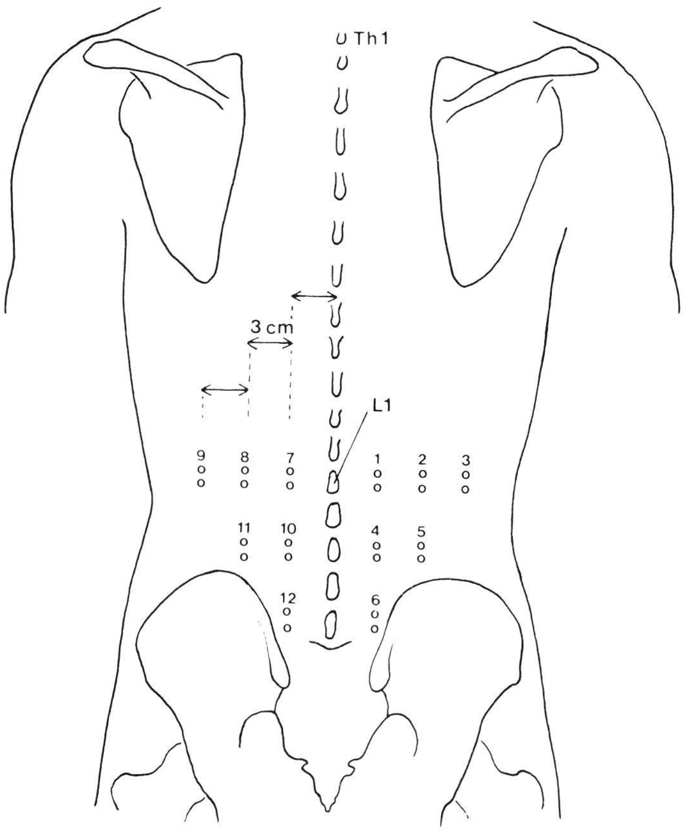

Fig. 3.1a. Position of the electrodes in the longitudinal arrangement. Three columns of the intrinsic lumbar back muscles are recorded: the medial (consisting mainly of the multifidus muscle) with electrode 1, 4 and 6 (7, 10 and 12), the intermediate (longissimus muscle) with electrode 2 and 5 (8 and 10) and the lateral (iliocostalis muscle) with electrode 3 (9).

were converted into digital signals with a sample frequency of 2000Hz for further processing on a PDP11/70 computer.

The shape of two EMG samples with a duration of 1 s was compared. By shifting one signal in time (-10 ms < t < 10 ms, with steps of 1 ms) the correlation between both EMG signals could be computed for every time lag. These values together yield the CCCF. The absolute maximum of this CCCF was computed to enable comparison of subjects. There is no rational basis

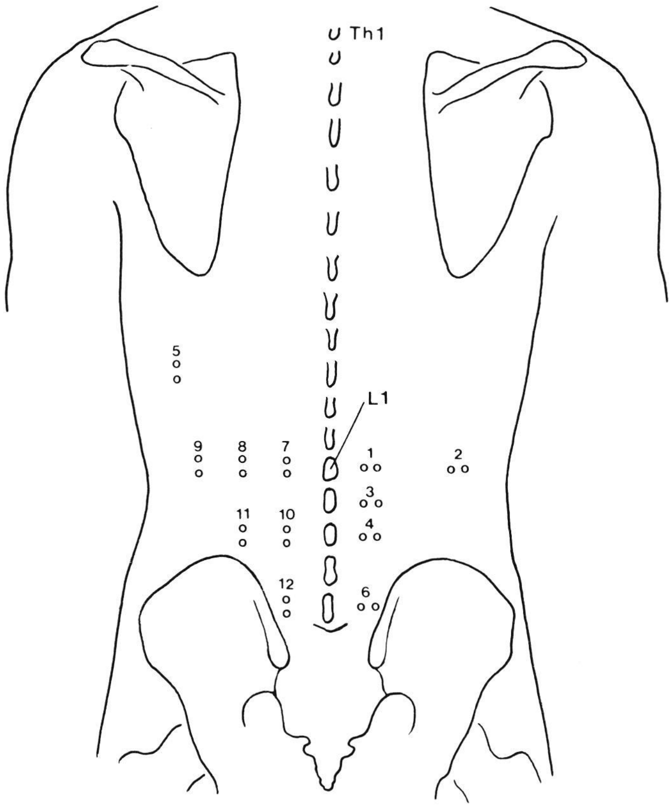

Fig. 3.1b. Position of the electrodes in the partly transverse arrangement.

for any negative correlation, so every maximum deviation from zero is taken as the maximal positive correlation.

In order to verify whether the CCCF can be used as a measure for the presence of cross-talk, different experiments were performed.

The influence of the channel acquisition time of the A.D.- convertor on the CCCF was tested. The same EMG was recorded simultaneously on all 12 channels (EMG from walking as well as in static situations). The CCCFs (in fact the "auto"-correlation coefficient functions) were computed between all channels.

To establish the influence of sources, other than the muscles, on the CCCF, one electrode (number 2) was replaced by making a short-circuit of both poles in one subject. The absolute maximum in CCCF was calculated between this signal, which contained all the instrumental and environmental noise and power-line-induced-a.c.-components, and the EMG of the ILBM (electrode 1, 3, 4 and 5; see fig. 3.1a). Recordings were made during an isometric contraction of 0, 10, 20, 30, 40, 50, 60, 70, 80, 90 and 100 % MVC (= maximum voluntary contraction). The experimental conditions are as described in chapter four. The amplification was unchanged between the different force levels. In doing so, we produced a wide variation in the signal to noise ratio.

The influence of the distance between electrodes on the absolute maxima in the CCCF was tested using the t-test ($p<0.05$). The effects of different muscle activities on the absolute maxima in the CCCFs were tested in pairs (sign test, $p<0.05$) in order to exclude interindividual differences. Each absolute maximum was compared with one in another experimental situation. For instance 48 paired comparisons were made between 10% and 60% MVC (8 subjects each of them having 6 pairs of electrodes located 30 mm apart, see table 3.1).

A. In experiment A, 8 subjects extended their backs against a resistance exerting 10 % and 60 % MVC of the ILBM in order to study the influence of the activity level on the CCCFs. At both activity levels EMG was amplified as much as possible in order to keep noise levels from the amplifiers and the tape relatively as low as possible.

B. In order to investigate the influence of activity of the latissimus dorsi muscle on the EMG of the ILBM a strong contraction of this muscle on the right side was combined with the exertion of 10 % and 60 % MVC of the ILBM and compared with experiment A (6 subjects). Latissimus dorsi activity might increase cross-talk (between electrode 2 and 3; 1 and 3; 3 and 5). Activation of the latissimus dorsi was imposed by adduction, endorotation and extension of the right arm against resistance.

C. ILBM-activity was recorded during walking on a treadmill (4.0 km/h) and compared with the results of experiment A (6 subjects). The cross-talk in this dynamic situation may differ from both foregoing static experiments, because of variation in activity levels or a displacement of the contracting muscles with respect to the surface electrodes.

D. In 8 subjects the three series of experiments A, B and C have been repeated to enlarge the number of paired comparisons resulting in series D(A10), D(A60), D(B10), D(B60) and D(C). Electrodes on the right side were transversely arranged (see fig. 3.1b) to detect any possible dipolar effect on the amount of cross-talk. Two electrodes on the right side were compared with two electrodes with equal distances on the left side. As space for only 11 electrodes was available in the transverse arrangement, one electrode was free for recording the left latissimus dorsi activity. This electrode (number 5) was placed 40 mm caudal to the inferior angle of the left scapula. The same position was chosen by Basmajian (1978) using wire electrodes. This enabled us to compute the CCCF between this muscle and the ILBM. The experiments of B were repeated, but instead of the right latissimus dorsi, the left was activated.

3.3 Results and discussion.

Although the channel acquisition time amounted to 117 μs, the mean absolute maximum in the "auto"-correlation coefficient functions of 12 leads was 0.996 (SD = 0.0071). The relatively small influence of the channel acquisition time is due to the dominance of low frequencies in the EMG from the ILBM (see fig. 3.2).

The absolute maximum in the CCCF between the signal from the short-circuited electrode and the EMG depends strongly on the EMG amplitude (see fig. 3.3). Above 10 % MVC CCCF values were below 0.1 and at low force levels (10 % MVC) high CCCF values were found, indicating the importance of a favourable signal to noise ratio.

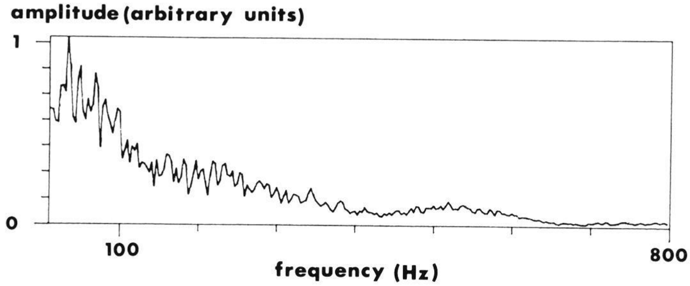

Fig. 3.2. The power spectrum of EMG derived with electrode 1 (see fig. 3.1a) exerting 10 % MVC, indicating that the mean power frequency is below 100 Hertz.

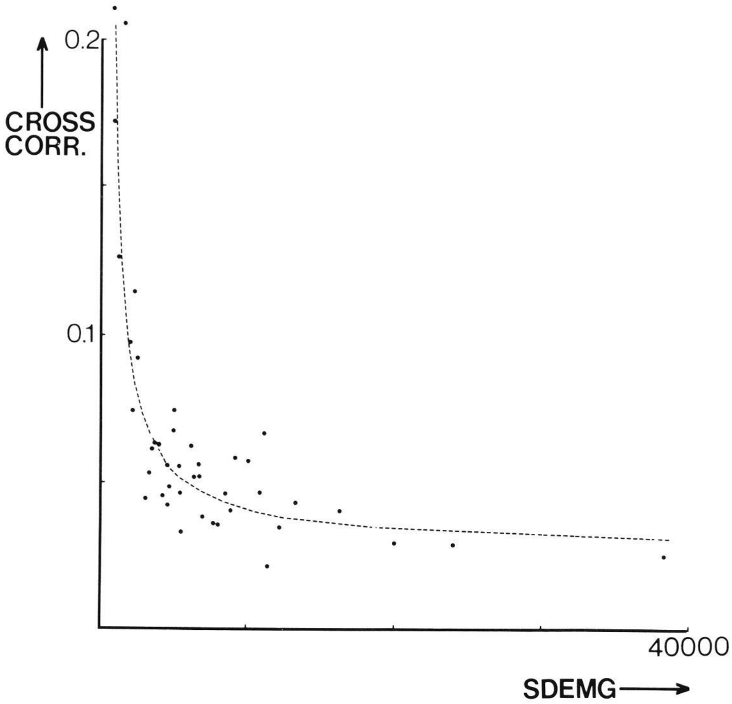

Fig. 3.3. The influence of the amplitude of EMG on cross-corre-lations of one subject. An increase in the standard deviation of the EMG (= SDEMG) describes the increased amplitude in the EMG. For larger SDEMG the cross-correlation of this EMG with a short-circuited electrode diminishes. With the least square method the best fitting curve (= dotted line) was found to be: $y = 131/(x + 135) + 0.030$ (corr.= 0.957). SDEMG and force have a relationship: SDEMG = 240F + 12 (corr. = 0.859; F in % of maximum force).

Table 3.1. Paired comparisons used to test the influence of different factors on the absolute maximum in the cross-correla-tion coefficient function (sign test, n= not significant, s= significant, p<0.05). Activities of the ILBM were recorded during: A=isometric contraction, B=isometric contraction with latissimus dorsi activity, C=treadmill walking, D= activities like A, B and C, but with a partly transverse arrangement of the electrodes.

influencing factor	paired experiments	number of subjects performing both exp.	mean of the absolute maxima with a distance between both electrodes of 30mm	60 or 70mm	85mm or more
force level	A10 % -A60 %	8	.56-.54n	.24-.26n	.14-.14n
	D(A10)-D(A60)	8	.53-.58n	.20-.21n	.15-.16n
	B10 % -B60 %	8	.50-.51n	.20-.22n	.16-.17n
	D(B10)-D(B60)	8	.61-.50n	.25-.21n	.17-.15n
			electrode nos. (see fig.3.1) 2-3	1-3	3-5
lat.dorsi activity	A10 % -B10 %	6	.59-.48n	.30-.24n	.25-.23n
	A60 % -B60 %	6	.57-.50n	.34-.30n	.32-.18n
			electrode nos. 8-9	7-9	9-11
	D(A10)-D(B10)	8	.58-.58n	.20-.30s	.24-.29n
	D(A60)-D(B60)	8	.48-.48n	.21-.12n	.19-.21n
isometric-treadmill	A10 % -C	6	.56-.41s	.24-.20n	.14-.16n
	A60 % -C	6	.54-.41s	.26-.20n	.14-.16n
	D(A10)-D(C)	8	.53-.36s	.20-.21n	.15-.16n
	D(A60)-D(C)	8	.58-.36s	.21-.21n	.16-.15n
longitudinal (left) versus transverse (right)	D(A10)L-R	8	.65-.45s	.19-.17n	.17-.19n
	D(A60)L-R	8	.53-.66n	.23-.28n	.16-.17n
	D(B10)L-R	8	.56-.68s	.25-.29n	.17-.17n
	D(B60)L-R	8	.54-.44n	.18-.31s	.16-.18n
	D(C) L-R	8	.38-.33n	.21-.23n	.17-.19n

The results of the cross-talk experiments A to D are summarized in table 3.1. The CCCFs show a strong relationship with the distance between two electrodes: the absolute maxima in CCCFs become higher with decreasing distances between two electrodes (see fig. 3.4 and table 3.1). Those electrodes located 30, 60 or 70, 85 and 90 mm or more respectively apart have a mean absolute maximum in CCCFs of 0.509 (SD = 0.106); 0.219 (SD = 0.060); 0.159 (SD = 0.039); and 0.148 (SD = 0.028) respectively. Except for electrodes located 85 and 90 mm (or more) apart, these maxima differed significantly from each other (t-test, p<0.05).

The CCCF values are not only influenced by the distance between the electrodes, but also by the biological material. The mean of the absolute maxima between electrodes located 60 mm apart on one side of the spine is 0.233 (SD = 0.064), whereas electrodes 60 mm apart crossing the spine show a mean of 0.179 (SD = 0.055). For a distance between electrodes of 85 mm these mean values were resp. 0.185 (SD = 0.036) and 0.145 (SD = 0.033).

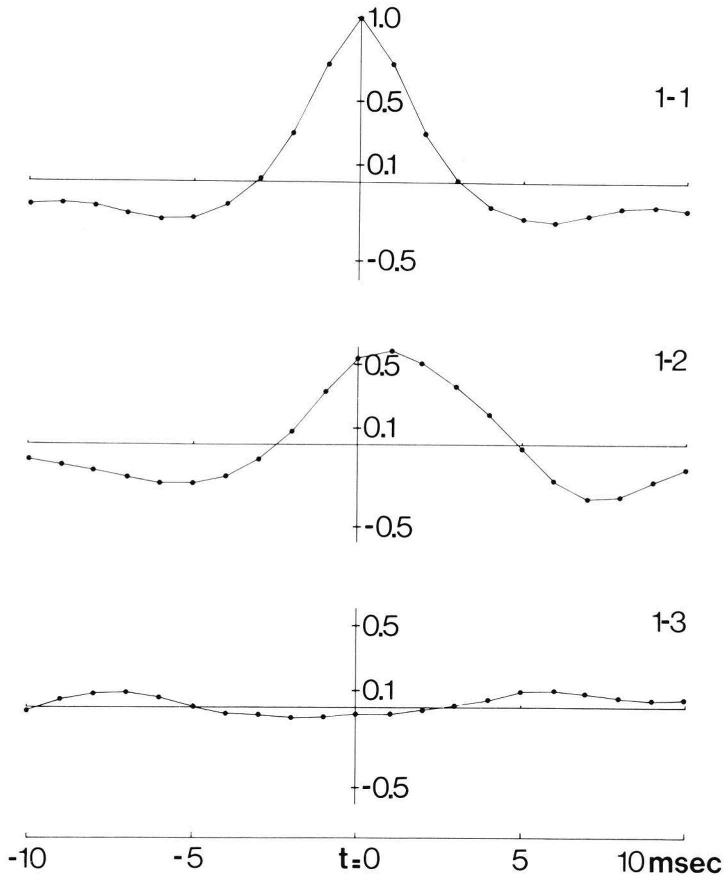

Fig. 3.4. The cross-correlation coefficient functions from -10 ms to 10 ms for subject F between electrode 1 and 2 (1-2), 1 and 3 (1-3) and the "auto"-correlation coefficient function for electrode 1 (1-1). 0 ms is marked with a vertical line.

These differences may be caused by a difference in resistance of the living tissue. Indeed, Geddes & Baker (1967) report a ten-fold difference in electrical resistance of bone and muscle.

The QRS complex of the ECG, is especially visible in the raw EMG-signals at L1 level, resulting in higher CCCF values at L1 level than at L3 or L5 level for electrodes located 60 mm or more apart (see table 3.1; A10 %, B10 %, D(A10) and D(B10)). Electrodes located 30mm or more apart are probably influenced more by crosstalk of the ILBM than by the ECG.

A. Producing 10 % and 60 % MVC.
Paired comparison of 8 subjects producing 10 % and 60 % MVC showed no significant differences (see table 3.1), when the EMG is amplified as much as possible. Thus, the activity level of the muscle is of minor importance in cross-talk, provided that the signal to noise ratio is favourable.

B. Producing 10 % and 60 % MVC with latissimus dorsi activity.
Latissimus dorsi activity is also of minor influence on the CCCF values. Even electrodes positioned on the ILBM close to the latissimus dorsi muscle (electrode 1, 2 and 3), showed no increase in CCCF values, compared with the experimental situation without latissimus dorsi activity (see table 3.1).

C. Walking 4.0 km/h.
The absolute maxima in the CCCFs with a distance between two electrodes of 30 mm, were significantly lower when walking than during an isometric contraction. This is reasonable because in our recording time of 1 s the EMG-signal varied during walking in contrast with isometric force exertion. Besides, during walking variation may occur in the distance between sources and electrodes due to the relative displacements between muscles and skin, resulting in lower absolute maxima.

The absolute maxima in the CCCFs, with a distance between two electrodes of 60 mm or more, were not significantly different between the experiments during walking and those during force exertion.

D. Transverse arrangement of electrodes.
Two electrodes positioned 30 mm apart in a transverse arrangement produced comparable high values to those in the longitudinal arrangement (see table 3.1). Comparing equal electrode distances on the left side of the back (longitudinal arrangement) with the right (transverse arrangement), we found that 3 out of 15 comparable positions showed significant differences in absolute maxima. However, as is shown in table 3.1, the CCCF values are more influenced by the distance between electrodes than by the direction of the dipole axis.

In all experiments, except for those with a provoked latissimus dorsi muscle activity, electrode 5 recorded little activity. Strong activity of the latissimus dorsi muscle (as in experiment D(B10) and D(B60)) resulted in an absolute maximum of 0.37 whilst exerting 10 % and 60 % MVC, when correlated with electrode 9. Thus, latissimus dorsi activity influences EMG of the iliocostalis, but only when it is strongly activated.

Two electrodes positioned more than 30 mm apart showed absolute maxima in the CCCFs less than 0.30. Although these values indicate the presence of cross-talk, this does not render EMG recordings unspecific. Even with the highest CCCF values of 0.70, some (about 50 %) information is stored which cannot be found in other leads.

3.4 Conclusion.

The CCCF is a simple technique to validate EMG-recordings and is therefore, indispensable in EMG investigations. When the raw EMG has been recorded over its full width of amplitude, CCCF values close to 0 are preferable. Nevertheless, one has to be alert that recorded EMG-signals also depend on the conduction characteristics of the tissues.

In recording the electromyographic activity of the ILBM, the electrode location described by Andersson et al. (1974) proved to be the most suitable. A more detailed EMG recording (using more than 12 surface electrodes) is not worthwhile, because cross-talk will increase. A less detailed recording is also not advised because it leads to loss of essential information from parts of the ILBM (Dofferhoff & Vink, 1984).

In the lateral upper region (L1) some cross-talk may be produced by a strongly activated latissimus dorsi muscle, which is not the case during walking.

The ECG was visible in the raw EMG at low force levels (<10 % MVC) especially at L1 level.

3.5 Summary

The cross-correlation coefficient functions (CCCFs) between twelve bipolar surface electrodes, placed symmetrically on the intrinsic lumbar back muscles (ILBM) were computed in order to estimate the amount of cross-talk. It was found that the CCCF values were mainly influenced by the distance between the electrodes. Other factors, such as static versus dynamic experimental conditions, the activity level, the angle between the dipole axis and the muscle fibers, the ECG, noise, power-lineinduced-a.c.-components and resistance of biological material, have less influence, but nevertheless they do change the CCCF values.

The absolute maximum in the CCCF can be used to validate EMG-signals. Absolute maxima in the CCCF within the range from 0 to 0.30 are considered as sufficient specific EMG-signals. The technique can be used to optimise the location of electrodes for the selective recording of localised muscle activity.

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4. A functional subdivision of the lumbar extensor musculature
Recruitment patterns and force-RA-EMG relationships under isometric conditions

P.Vink, E.A.van der Velde & A.J.Verbout

Electromyography and Clinical Neurophysiology 28: 517-525, 1987

4.1 Introduction

The function of the ILBM (=intrinsic lumbar back muscle) as indicated by their electromyographic signal, has been described often in the literature (Floyd & Silver, 1955; Portnoy & Morin, 1956; Morris et al., 1962; Pauly, 1966; Lucas, 1970; Jonsson, 1970 (a); Donish & Basmajian, 1972; Örtengren & Andersson, 1977; Valencia & Munro, 1985). Although authors conclude that ILBM activity is dependent on the position of the body center of gravity and counteracts the gravity force, their exact relationship (the force to EMG relationship =FER) is still unknown.

Grieve & Pheasant (1976) studying 2 subjects and using only 2 bipolar surface electrodes, one on each side of the spine, found a curvilinear relationship between a computed torque and the rectified and smoothed EMG. Andersson et al. (1980) on the contrary, described a linear relationship between calculated loads on the lumbar spine and the mean myoelectric signal amplitude, averaged over 10 subjects.

For other muscles than the ILBM FERs have been extensively studied. An excellent review is given by Bouisset (1973) and supplemented by Perry & Bekey (1981). Different FERs have been described. This may be due to a variety of reasons, such as certain aspects of the measuring methods (the distance between electrodes (Möller, 1966), technical conditions (De Vries, 1968), electrode type, sampling rates, noise, filtering and amplification, rectification and averaging) or of the physiological conditions (the proper and actual length of the muscle (Grieve & Pheasant, 1976; Hof & v.d. Berg, 1977; Hof, 1980) and fatigue). Thus some authors found a linear FER under isometric conditions (Lippold, 1952; Inman et al., 1952; De Vries, 1968; Hof

& v.d. Berg, 1977; Antti, 1977), whereas others found a curvilinear FER (Möller, 1966; Vredenbregt & Rau, 1973; Antti, 1977).

Studies in which the different individual columns of the ILBM have been investigated are few. Klausen (1965), who investigated the form and function of the loaded spine, concluded that the short, deep intrinsic muscles of the back play an important role in stabilising the individual joints. The long back muscles and the abdominal muscles are responsible for the stabilisation of the spine as a whole. In accordance with this view Donisch & Basmajian (1972) stated that the transversospinal muscles adjust small movements between individual vertebrae, while movements of the whole spine were supposed to be performed by muscles with more favourable mechanical properties for these movements.

Based on previous studies indicating the location of the different columns of the ILBM (Reichmann & Jonsson, 1967; Jonsson, 1970 a), Jonsson (1970 b) could distinguish three columns of the ILBM with different functions in a EMG-study using wire electrodes. These three parts sometimes show even different activities at different lumbar levels (Jonsson, 1970 a). Extension and flexion evoke activity in the whole ILBM, whereas lateroflexion and rotation of the trunk especially evoke activity of the m. iliocostalis and m. longissimus.

The purpose of our investigation is to establish the different functions in stabilising the spine of the three columns of the ILBM. The ILBM can be subdivided into a medial column (mainly made up by the multifidus, spinalis and the longissimus muscle), an intermediate column (longissimus muscle) and a lateral column (iliocostalis lumborum muscle). In this experiment we studied the electromyographical differences between the columns of the ILBM in one static position, varying only the extension force. In later studies different positions and dynamic situations will be included as well. In a pilot study (Vink, 1986) evidence was found that the relationship between an external force and EMG differed for medial and lateral parts of the ILBM. In order to study these differences more in detail two series of experiments were performed. In the first series the relationship between force and EMG for the ILBM is calculated using 10 subjects and 5 forces. In the second series the experiment is done with 5 subjects using 10 forces with sequential changes to see whether the sequence in the experimental procedure would influence this relationship.

4.2 Materials & Methods.

All 15 subjects (see table 4.1) had no history of back disease and were healthy students, who practiced sports regularly.

Table 4.1. Data of the 15 subjects. In subject A to J the first series of experiments and in subject K to O the second series were done.

subject	weight (kg)	length (m)	maximum force (N)	age	male/ female	regular sport
A	81	1.89	735	39	m	+
B	68	1.69	784	24	m	+
C	74	1.81	617	29	m	+
D	68	1.74	833	34	m	+
E	60	1.72	314	21	f	-
F	54	1.69	343	22	f	+
G	63	1.83	706	19	m	+
H	75	1.76	647	27	m	+
I	59	1.59	637	18	f	+
J	54	1.68	353	22	f	+
K	70	1.70	1030	24	m	+
L	72	1.91	637	20	m	+
M	57	1.64	441	20	f	+
N	73	1.80	686	30	m	+
O	72	1.83	588	25	m	+

The subjects stood with extended knees, their arms along the body and the pelvis and thighs against a board to push off (see fig. 4.1). In this position they extended their back against a mechanical resistance to produce an as nearly isometric action of the ILBM as could be achieved. The extension force was measured externally with a Salter spring balance (gauged accuracy better than 5N, maximum force 980 N), which had been fixed at a 3 cm wide girdle around chest and arms with its upper edge approximately at the level of vertebra Th 5 and just below the greater tubercles of the humeri.

The girdle was adjustable for every subject to ensure that the body had a vertical position while the girdle was kept in a horizontal position. This position appeared to be the most comfortable for the subjects and resulted in very reproducible force values for the maximal voluntary contraction (=MVC). With the girdle at a lower level around the chest (approximately Th11) extension became very uncomfortable and the force output was strongly influenced by respiration motions. As a result the FER showed a large variation in EMG values for the same forces.

Myoelectric activity was recorded using bipolar surface electrodes. These were gold-plated home-made bipolar electrodes with a diameter of 7.5 mm

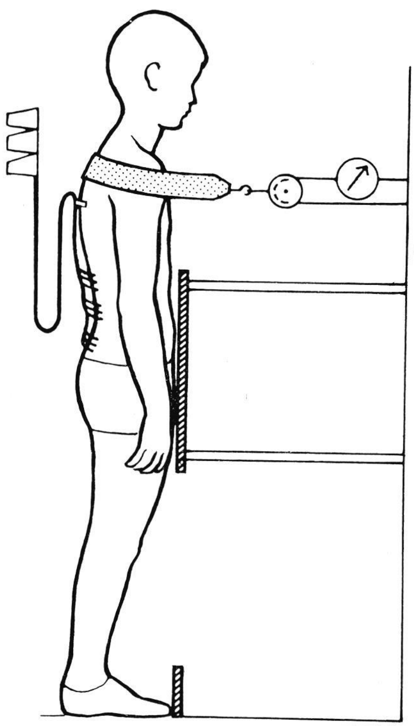

Fig.4.1 The experimental set-up, in this standing position the subject exerted different external forces under nearly isometric conditions. The pulley was only used for one subject (K).

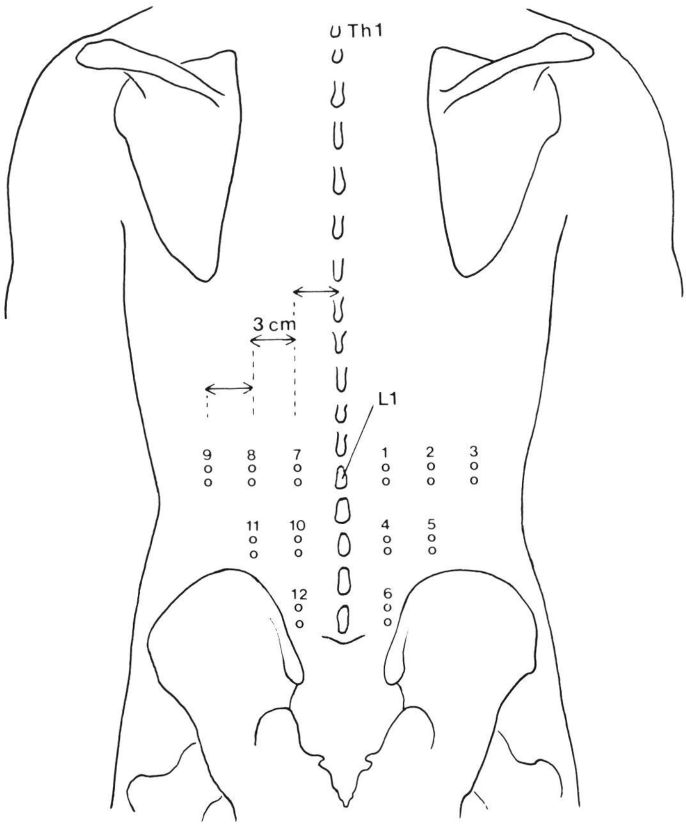

Fig.4.2 The position of the 12 electrodes.

and a bipolar center to center distance of 21.5 mm. Twelve electrodes were applied bilaterally in a longitudinal bipolar orientation at levels L1 (three electrodes), L3 (two) and L5 (one), using a location that enabled us to compare our results with those of Andersson et al. (1974) and Jonsson (1970 a) (see fig. 4.2).

At L1 two electrodes were placed at resp. 3 cm and 6 cm laterally to the spinal processes whereas a most lateral electrode was placed 1 cm medial to the lateral border of the ILBM (8-9 cm lateral to the spinal processes). At level L3 two electrodes were placed at resp. 3 cm and 6 cm and at level L5 one electrode was placed 3 cm laterally to the spinal processes.

The strongly curved skin surface in the L5 area required a pair of two separate electrodes (Hellige silver disc electrodes; type 21717802; with an internal diameter of 6 mm; with bipolar distance from center to center of 21 mm). The 12 signals of the ILBM were preamplified with a fixed gain of 100 (the differential amplifier had an input impedance of 100 M , a cmmr > 20.000 and a frequency response being flat between 1 and 10^5 Hz) and led further by a shielded cable to the main amplifier with variable gain up to 1000.

Recordings lasted 2 or 3 seconds (vide infra) and were stored on magnetic tape. Afterwards 3 samples of rectified and averaged EMG (RA-EMG) of 1 second, indicated by the display of the integrator were taken from each run. Of these 3 samples the highest value was noted and the procedure was repeated. The mean of four maximum values was used in the calculation of the FER for each electrode, for each force and for each subject. The forces and RA-EMGs were expressed as percentages of the maximum RA-EMG. The latter (used as a reference) was calculated for each run and each electrode at the 100 % MVC force level. The FER was estimated with a regression curve using an equation of Kadefors (1978), which we modified:

$$E = k\,F^{q} + c$$

(E=RA-EMG in % of maximal RA-EMG, F=force output in % of maximal force; k, q and c are constants for which $k = 100^{1-q}$ holds, due to normalization). In our modification the constant c was added because the ILBM are not fully silent in an upright standing subject exerting minimal force on the spring balance. The values of k, q and c were calculated for each electrode with the least squares estimation. That means that q is the value for which F and the observed E had the highest correlation coefficient.

In all cases the correlation coefficient turned out to vary between 0.866 and 1.000. The differences between the power of the curves of the medial, intermediate and lateral electrode were analyzed with the distribution free test of Friedman or where appropriate with Wilcoxon's signed-rank test.

4.2.1 The first series of experiments.

Ten subjects had to exert force at 5 levels during which EMG was recorded. Firstly a MVC of 2 seconds delivered the reference level of maximal force, which was used also for adjustment of the amplification of the EMG to the band width of the recorder. Secondly forces of 10 %, 25 %, 50 %, 75 % and 100 % MVC were exerted during 3 seconds with resting intervals of about 5 seconds. For adjustment subjects were able to read off their force exertion.

4.2.2 The second series of experiments.

For 5 subjects a second series of three different runs was carried out.

I The subject had to pull 10 %, 20 %, 30 %, 40 %, 50 %, 60 %, 70 %, 80 %, 90 % and 100 % of the MVC during 3 seconds with resting intervals of about 5 seconds.

II After a resting period of 15 minutes the subject had to exert the same forces for 3 seconds in a reversed sequence (decreasing values) again with resting intervals of 5 seconds.

III Again after a resting period of 15 minutes the subject had to exert the same forces as in run I (increasing values), however with resting intervals of 60 seconds.

In this series the number of measuring points of the first run was increased while the signal to noise ratio was improved for the following reason. In the first series we noticed a great difference between the RA-EMG values of 10 % and 100 % MVC. This had an unfavourable effect on the signal to noise ratio in the recordings at lower force levels. Therefore for these force levels (10-50 % MVC) the EMG-signal was amplified 10 times more in the second series.

The second and third runs may be helpful to detect effects of fatigue. Supposing that fatigue might produce a greater synchronization of potentials (Moretani & De Vries, 1978) this could be reflected by an increased amplitude of the RA-EMG. If this is true 100 % MVC at the first force level of run II would produce the lowest RA-EMG values, while higher values can be expected at the end of run III and highest at the end of run I. Due to normalization the FER-curve of run II would run above the curve of run I with the curve of run III between them.

4.3 Results.

The FER was determined 297 times (120 in the first and 177 - three times the electrodes became unstock or exceeded the range of the recorder - in the second series of experiments). In 124 out of 297 the FER appeared to be close to linear ($0.5<q<1.5$); especially this holds for the medially placed electrodes (104 out of 150; in the first series 34 out of 60 (57 %) and in the second series 70 out of 90 (78 %). For the intermediate and lateral electrodes however a close to linear FER was found only in 20 out of 147 registrations (in the first series 5 out of 60 (8 %) and in the second series 15 out of 87 (17 %)). For these laterally placed electrodes the RA-EMG increased more than linear with increasing forces.

For 141 FER-curves a significant difference was found between the 3 columns of the ILBM at L1 level (Friedman's test, $p< 0.05$). In almost all cases the curves of the lateral electrodes had a higher power than those of the intermediate, whereas the latter had higher powers than the curves of the medial electrodes.

Table 4.2. Example of the FER (External force and RA-EMG relationship) for some electrodes of subject H. The FER was computed by calculating the 5 points in which 10 %, 25 %, 50 %, 75 % and 100 % MVC was exerted.

Pair of electrodes	FER ($E= k\,F^{q} + c$)	correlation-coefficient	power(q)	p.value
1	$E=1.04 \times F^{1.1} - 2.31$	0.995	1.1	0.0004
2	$E=0.97 \times F^{3.7} + 2.73$	0.999	3.7	0.00002
3	$E=0.99 \times F^{4.9} + 0.53$	0.995	4.9	0.000006
4	$E=0.92 \times F^{1.0} + 5.96$	0.995	1.0	0.0004
5	$E=1.02 \times F^{1.6} - 1.57$	1.000	1.6	0.000004
6	$E=1.02 \times F^{2.3} - 1.85$	0.999	2.3	0.00002

Table 4.3. q values of the 10 subjects. This concerns the first series of experiments exerting 10 %, 25 %, 50 %, 75 % and 100 % MVC.

	electrodenumbers											
subject	1	7	4	10	6	12	2	8	5	11	3	9
A	1.8	1.6	1.7	1.6	1.8	1.5	1.7	3.1	2.6	2.6	3.2	4.3
B	1.6	2.0	2.0	2.0	2.0	1.6	2.9	2.2	2.7	2.6	3.1	3.1
C	0.9	1.0	1.0	1.0	1.0	1.3	1.6	2.0	1.6	1.9	2.1	2.9
D	1.6	4.1	1.0	1.6	0.5	1.1	3.0	2.7	2.5	2.5	4.0	4.1
E	1.6	1.5	1.3	1.4	1.4	2.7	2.9	2.3	1.4	4.0	4.9	4.9
F	1.2	1.1	1.1	1.2	1.0	0.6	1.8	1.5	1.8	1.3	2.8	2.5
G	1.3	1.6	0.9	0.7	1.3	1.0	1.3	1.9	3.1	1.8	2.6	2.6
H	1.1	1.1	1.0	1.5	2.3	1.7	3.7	1.9	1.6	0.9	4.9	1.9
I	1.1	1.7	1.7	1.4	1.3	1.1	3.2	3.7	1.4	1.9	3.9	4.2
J	1.1	0.8	1.6	1.6	0.9	1.0	3.2	1.9	1.7	1.5	3.2	3.2

Table 4.4. q values of the 5 subjects this concerns the second series of experiments exerting increasing forces of 10 %, 20 %, 30 %, 40 %, 50 %, 60 %, 70 %, 80 %, 90 % and 100 % MVC for run I, decreasing forces from 100 to 10 % for run II and increasing forces with resting intervals of 1 minute for run III. Three times q could not be computed due to technical problems (-).

Subject	electrodenumbers											
exp	1	7	4	10	6	12	2	8	5	11	3	9
KI	1.8	1.9	1.3	0.6	4.4	2.5	2.4	1.5	1.7	1.9	2.1	1.7
KII	1.6	1.4	1.8	1.5	2.5	2.1	2.0	1.4	2.4	1.5	1.9	2.4
KIII	1.2	1.0	1.3	1.1	2.5	2.8	2.1	1.5	1.7	1.9	1.8	3.4
LI	0.6	1.0	1.4	1.7	0.9	1.5	2.1	3.1	1.6	2.0	2.2	4.0
LII	0.7	0.6	1.3	1.2	1.0	1.2	1.4	1.5	1.1	1.2	1.6	2.1
LIII	0.5	0.5	1.6	1.4	0.7	1.0	2.2	1.7	1.3	1.5	1.8	2.7
MI	1.1	1.5	1.1	1.1	0.5	0.5	2.6	2.1	1.3	2.0	3.4	4.1
MII	1.0	1.3	1.7	1.3	0.8	0.7	3.2	-	1.9	1.6	4.8	2.6
MIII	0.9	0.7	1.5	1.0	0.5	0.9	3.0	-	1.5	1.5	3.9	3.6
NI	0.8	0.8	0.8	1.0	1.0	0.9	1.4	1.5	1.3	1.0	1.8	1.3
NII	0.8	1.0	0.9	1.3	0.7	0.9	1.5	1.4	1.0	1.9	1.8	2.5
NIII	0.8	0.9	0.8	1.2	0.8	1.1	1.4	1.3	0.9	1.3	1.9	1.7
OI	1.0	2.1	1.0	1.1	1.0	1.0	1.5	2.1	2.1	1.9	2.7	2.9
OII	1.0	2.1	1.5	1.1	1.1	1.1	1.9	2.7	2.1	1.8	3.2	3.4
OIII	1.2	1.4	1.1	0.9	1.0	1.0	1.5	-	2.1	1.8	2.6	2.8

Concerning the L3 level, the Wilcoxon signed rank-test ($p<0.05$) for paired observations applied to 100 FERs showed a significantly higher power for the intermediate electrodes than for the medial ones.

Because in the two series of experiments the steps of the produced forces were different, the other results of the two series will be presented separately.

4.3.1 The first series of experiments.

Of 120 FERs in 39 (33 %) the power varied from 0.5 to 1.5, in 43 (36 %) the power varied from 1.5 to 2.5, and in 38 (32 %) the power was above 2.5. The mean power for the medial electrodes was 1.40, for the intermediate 2.25 and for the lateral 3.42 (see fig. 4.3 and table 4.2). Comparing the

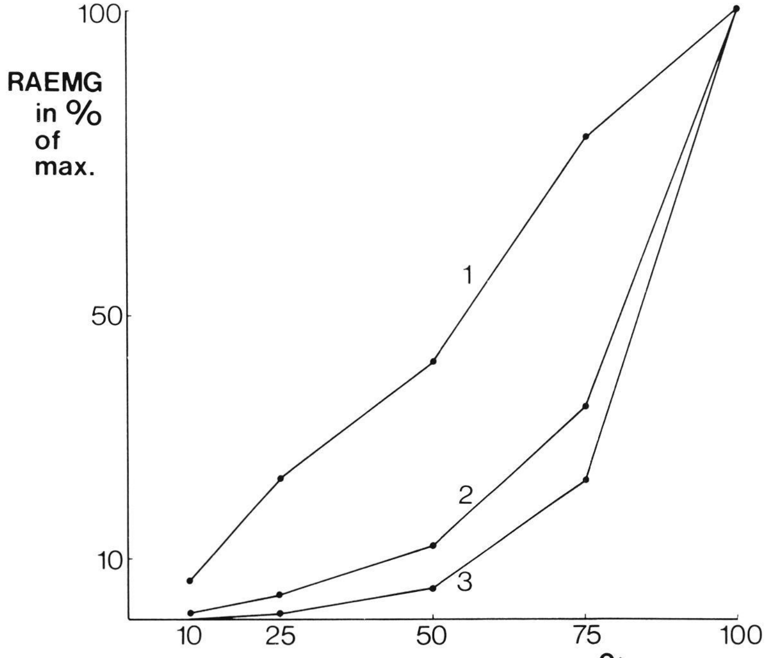

Fig.4.3 The FER-curves computed for the first series of experi-ments of subject H. The force and RA-EMG are both normalized in percentage of the maximum. For the individual electrodes (the number of the curve corresponds to the electrodenumber in fig. 4.1) a different relationship is shown.

corresponding columns with each other at different levels only the electrodes of the intermediate column at L1 and L3 showed a significant difference (t-test: $P<0.05$): the mean power of the FER was for the L1 and L3 level respectively 2.43 and 2.07.

In these series high correlation coefficients ($0.990<r<1.000$) were found. Probably it is caused by the rather limited use of only 5 forces. Therefore the number of force levels was enlarged to 10 in the second series of experiments.

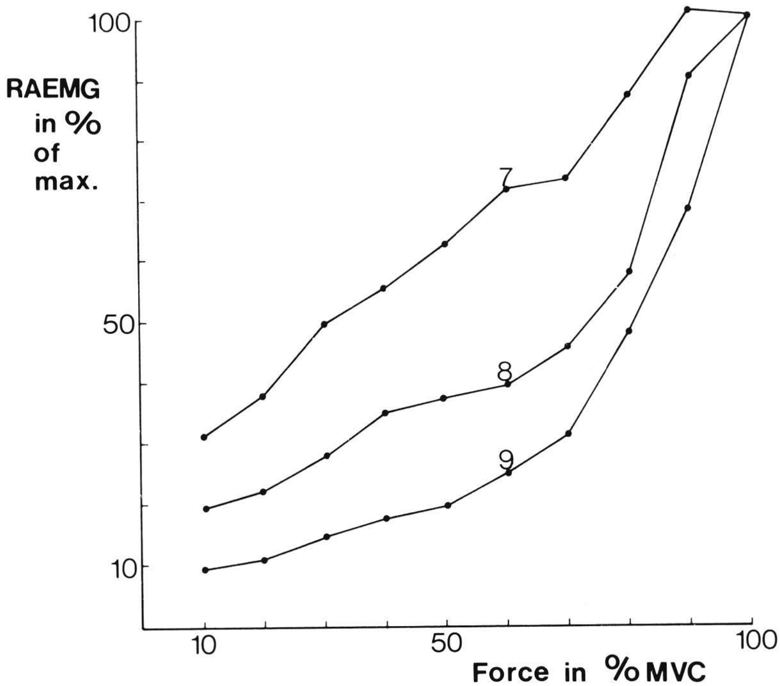

Fig.4.4 The FER-curves computed for the second series of experiments of subject L. It concerns run II with decreasing forces. The force and RA-EMG are both normalized in percentage of the maximum. The number of the curve corresponds to the electrodenumber in fig.4.1. For curve 7 the FER fitted with a correlation coefficient of 0.982 to equation $E=8.52F^{0.55}-7.4$*; this was for curve 8 and 9 resp.* $E=1.03F^{1.5}+9.7$ *(corr. 0.978);* $E=1.01F^{2}+6.1$ *(corr. 0.989).*

4.3.2 Results of the second series of experiments.

In the second series a similar difference between the values of the mean power of the FERs for the medial, intermediate and lateral electrodes was found: resp. 1.33, 2.79 and 3.55 (see table 4.3 and fig. 4.4).

In 85 out of 177 FERs (48 %) the power varied from 0.5 to 1.5; this variance concerned 78 % of the medial electrodes (70 out of 90). In 66 (37 %) it varied from 1.5 to 2.5, which applied to 65 % (37 out of 57) of the intermediate electrodes. In 26 out of 177 FERs (15 %) the power was above 2.5, this concerned 53 % (16 out of 30) of the lateral electrodes.

Compared with the first series powers were somewhat lower in the second series, although similar differences between the lateral and medial electrodes were observed.

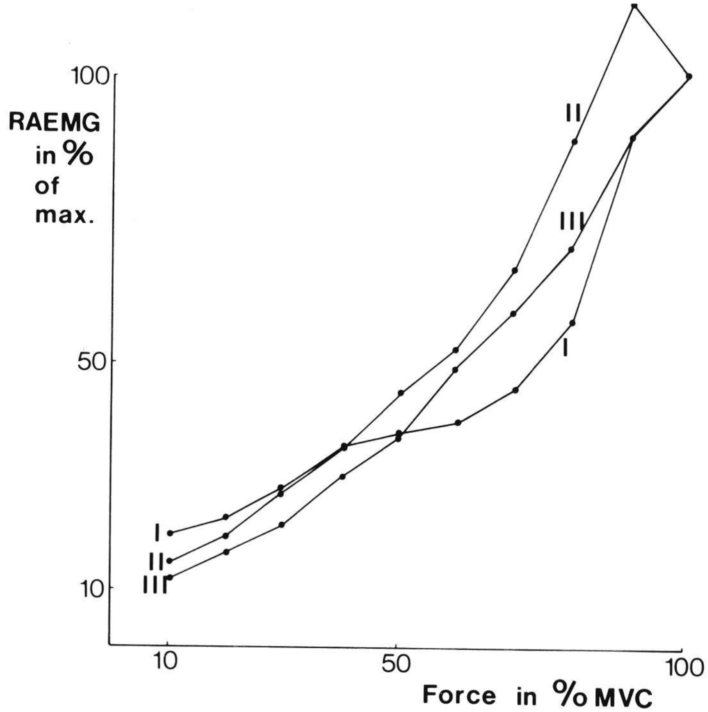

Fig.4.5 The FER-curves for electrodenumber 7 computed for the second series concerning run I, II and III (see text) of subject L. The relationships are described mathematically as, I, $E=0.88F^{.98}+22.15$ (corr. 0.992); II, $E=8.52F^{.55}$ -7.4 (corr. 0.982); III, $E=11.3F^{.5}$ -10 (corr. 0.992).

When the equation $E= k F^{q} + c$ is used, differences between the 3 runs of one subject appeared to be small or absent (see table 4.4), however differences between several subjects were greater.

The correlation coefficient varied in the second series between 0.866 and 1.000. These values are lower than in the first series, which is in agreement with the assumption that the use of 10 measuring points instead of 5 generally decreases the correlation coefficient.

The RA-EMG values of 100 % MVC for the three runs showed no significant differences (Friedman's test, $p<0.05$). For one subject (L) the empirically found FER curves of run I, II and III appeared to be distinguishable above 50 % MVC. This difference could not be demonstrated with the modified equation of Kadefors (1978). Ten of the 12 FER curves of this subject demonstrate a pattern in which the curve of run I was the lowest and the curve of run II the upper one with the curve of run III in between (see fig. 4.5). This phenomenon may be explained by fatigue (see discussion).

4.4 Discussion

During upright standing gravity's moment arms are relatively short and counteraction of the ILBM requires low activity levels. Indeed our results show a low and variable level of RA-EMG, this may be in accordance with the small shifts in posture during upright standing postulated by Ortengren and Andersson (1977). Apart from the variability in low levels of EMG the QRS complex of the ECG was also detectible. Thus the RA-EMG values under 10 % MVC were not taken into account in the FER.

Values above 10 % MVC could be influenced by fatigue. Fatigue may shift the FER upward (i.e. a higher RA-EMG for the same force) as experimentation proceeds (Moretani & De Vries, 1978). In our results fatigue had little influence. Only in subject L a possible influence of fatigue was seen. The 100 % MVC of run I produced higher RA-EMG levels than in the runs II and III for all 12 electrodes. A difference between the 100 % MVC levels of run II and run III was not observed. Probably the subject recovered within 60 seconds. The observed differences between the curves of subject L (see results) may be also explained by fatigue. Starting with high forces (run II) the subject is more fatigued than starting with low forces (run III) and due to the normalization procedure, FERs of increasing forces and short resting intervals were lowest.

Our results resembled to those of Grieve and Pheasant (1976) and Andersson et al. (1980). Grieve and Pheasant (1976) reported a curvilinear FER using two electrodes positioned on the ILBM. However, our correspon-ding electrodes showed a more regular increase in RA-EMG, probably because they calculated the torque with estimated magnitudes of the momentarms and gravity forces, while we measured force directly. However a problem in our experiment is the unknown relationship between external force (or torque) as measured with the girdle and the force exerted by the individual columns of the ILBM. The regular increase of the RA-EMG we observed may suggest a reproducible contribution of the more cranial intrinsic back muscles. Unfortunately a lower position of the girdle measuring more directly the extension force of the ILBM resulted in a much wider variation of the EMG-values, which were also lower as a consequence of the lower torque to be delivered. Respiratory movements, might also cause a wider variation resulting in a less regular increase of the RA-EMG values. Finally this lower position proved to be very uncomfortable for the experimental subjects interfering with a regularly increasing dosage of the exerted extension forces.

Nevertheless, the different columns of the ILBM showed similar differences between the powers of the mathematically represented FER curves as were found in higher positions of the girdle. Andersson et al. (1980) also estimated the gravity torque. Their values had a more regular increase because these were the averages of four electrodes in ten subjects. But such a procedure neglects a subdivision of the ILBM and obscures its possible functional differences. In our experimental set-up the gravity force was almost eliminated, as the position of the subject was upright. This appeared to result in a regular increase of the RA-EMG for every electrode and every subject.

A problem in recording muscle activity with EMG is that cross-talk plays a role, which means that also activity is recorded from muscles other than for which the electrodes are meant. It is shown (see chapter 3) that cross-talk is mainly present in two horizontally neighbouring electrodes.

Hypothetically it is possible that all the FERs are in fact linear and that their curvilinearity is an effect of cross-talk. If recruitment of the intermediate and lateral columns starts at higher force levels with a linear increase cross-talk may raise their RA-EMG values in such a way that curvilinearity is simulated.

At the other hand assuming that cross-talk is negligible and that the load sharing between muscles depends on the type of motor-unit (Dul et al., 1985) the force of two separate columns of the ILBM at L3 level can be estimated. The technique of Dul et al. (1985) calculates the load sharing between synergists starting from the hypothesis that muscular fatigue is minimized. So muscles with the greatest endurance, e.g. the muscles with the greatest percentage of type I fibers are recruited first. Širca and Kostevc (1985) found a statistically significant difference between the medial part at L3 level containing 63 % type I fibers and the intermediate part with 57 %. Application of the model of Dul et al. resulted in q values for the medial and intermediate part of resp. 0.9 and 1.1. However, these values as well as the difference between them are lower than the values we reported in our results. The EMG differences might even be greater when cross-talk is absent. Therefore the difference in recruitment pattern can not be explained completely by the difference in percentage of slow twitch fibers.

Referring to the results of Woittiez (1984) one may suppose that muscles with a relatively large number of short fibers are recruited first around rest length. These muscles have a relatively large physiological cross-section with a small range of activity and a high maximum force. The medial columns have probably the greatest number of short parallel fibers, while the intermediate and lateral columns have longer fibers. The momentarms of the three columns of the ILBM are almost equal in extension. Therefore it may be more efficient to recruit the part with the highest maximum force first.

Finally, our results seem to be corroborated by those of Jonsson (1970 a) who found different functions for the 3 columns of the ILBM: the lateral muscles being recruited later during an increasing loading of the spine within the sagittal plane. He found comparable results, notwithstanding the fact that he used wire electrodes and measured in several postures of flexion and extension.

Jonsson (1970 a) found also that the iliocostalis lumborum muscle was more concerned with activities in the frontal plane, which is not surprising in view of its favourable leverarm. We observed a similar phenomenon during walking (Dofferhoff & Vink, 1985).

These results suggest that the medial part of the ILBM is mainly active in maintaining posture, especially counteracting forces in the sagittal plane, while the lateral columns are more concerned with forces in the frontal plane, being recruited only for movements in the sagittal plane as additional muscles at higher and highest force levels.

4.5 Summary

The relationship between an external force and surface RA-EMG (rectified and averaged) was investigated for the intrinsic lumbar back muscles (=ILBM) in the upright position under isometric conditions. Of three columns of the ILBM the signals were recorded: the medial, the intermediate and lateral column which topographically corresponds grossly with resp. the multifidus, longissimus and iliocostalis muscles. For the three columns the relationship between an external force and RA-EMG (FER) differed significantly. The medial muscles showed a FER close to linear, the intermediate muscles had a curvilinearity with a higher power and the lateral muscles had the highest power. This means that the medial columns increase their activity proportionally during increasing force exertion whereas the lateral columns increase their activity mainly or start force deliverance above 50 % of the maximum forces exertion.

Our interpretation of this finding is that the medial part of the ILBM is mainly active in maintaining posture, and especially when counteracting forces in the sagittal plane. The lateral part is more concerned with forces in the frontal plane.

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5. Low back muscle activity and pelvic rotation during walking.

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Anatomy and Embryology 178: 455-460, 1988.

5.1 Introduction.

Little data are available about the functions of the intrinsic lumbar back muscles (=ILBM) during walking. Most EMG-studies describe the duration of the muscle activity of a small part of the ILBM and in a limited number of steps (Battey & Joseph, 1966; Waters & Morris, 1972; Dofferhoff & Vink, 1984). Studies of rotation of the pelvis are also few and concern a limited number of steps (Belenky, 1973; Inman et al., 1981; Thurston, 1982). In this study we try to extend the knowledge about ILBM activity and pelvic rotation recorded simultaneously during walking.

In an early study (Dofferhoff & Vink, 1984) a difference in activity time between lateral and medial parts of the ILBM was shown, whilst in a later study we were able to demonstrate different recruitment patterns of three columns of the ILBM in a static activity of the back (Vink, 1986a; Vink et al., 1987), but differences in its functions during walking are unknown.

5.1.1 Morphology of the ILBM

Under the thoracolumbar fascia two large muscles are found: the multifidus and the erector spinae muscle. The description of these muscles in standard textbooks of anatomy (e.g. Braus, 1954; Warwick & Williams, 1973) has recently been elaborated. The erector spinae muscle consists of two muscles, separated by the lumbar intermuscular aponeurosis (Bogduk, 1980). The longissimus muscle arises medially to the lumbar intermuscular aponeurosis and from the ilium. It inserts into the tips of the transverse processes of vertebrae Th4 to L4 (Bogduk, 1980). The iliocostalis muscle arises laterally from the lumbar intermuscular aponeurosis and inserts into the transverse processes of the first four lumbar vertebrae and is attached by slender flattened tendons to the angles of the inferior ten ribs (Bustami, 1986). The fibers of the iliocostalis muscle extend from caudomedial to rostrolateral and cross the fibers of the longissimus muscle laterally and posteriorly, which extend deep from caudolateral to rostromedial (Bustami, 1986).

The multifidus muscle, which is situated medially to the erector spinae muscles, consists of five bands, each extending from a lumbar processus spinosus to the sacrum (Macintosh et al., 1986).

5.1.2 Function of the ILBM during walking

The activity of the ILBM has been discussed previously (Dofferhoff & Vink, 1984). Because heel strike has a decelerating effect on the pelvis connected with the deceleration of the limb, the moment of inertia causes flexion of the upper part of the trunk with respect to the lower part. The bilateral activity of the ILBM is needed to counteract this trunk flexion. After heel strike unilateral ILBM-activity is needed to counteract the acceleration of the trunk in a contralateral direction (lateroflexion). In the present study which part of the ILBM exerts these two forces will be established. It may be expected that in this ribless region of the trunk the ILBM have an important share in the maintenance of dynamic equilibrium.

Cappozzo (1983) quantified the forces to be exerted by the ILBM. He calculated the intersegmental force and moment exchange between upper and lower body across a transverse section through the fourth lumbar vertebrae. He used 3-D kinematic data of head, upper limbs and upper torso from stereo-photogrammetry with the relevant inertial parameters, obtained from anthropometric measurements and estimation techniques provided in the literature. He estimated that the principal muscular moment on the fourth lumbar vertebra is in the sagittal plane. The extensor muscles should deliver a force peak either starting just before or just after heel strike to counteract the moment of gravity and inertia. The force peak ends at about one third of the subsequent stance phase.

The magnitude of the moment in the frontal plane is approximately one third smaller than the magnitude of the moment in the sagittal plane. Moments in the frontal plane reach a peak value at toe off. Since both the moment in the frontal as well as in the sagittal plane reach a peak simultaneously, Cappozzo (1983) predicts that the right ILBM may be expected to exhibit a major burst at right toe off and the left at left toe off.

The major moment in the transversal plane is approximately one fifth of the maximum moment in the sagittal plane, and mechanical activity of the ILBM is predicted to be small or absent during walking (Cappozzo, 1983).

In summary, the whole ILBM should exert force from just before heel strike until about one third of the subsequent stance phase to compensate for the moment in the sagittal plane. At right toe off the right ILBM, and at left toe off the left ILBM, are predicted to add a force to compensate for the moment in the frontal plane.

This is affirmed by a previous study (Dofferhoff & Vink, 1984) for the activity time. In this study it is investigated as to how these forces are exerted by the different parts of the ILBM. The activity time and the amplitude of EMG have been recorded together with stride times and rotations of the pelvis, in order to establish relationships between the activity of parts of the ILBM and pelvic rotations in different planes. It is probable that the pelvic position influences the gravity torque on the spine and, assuming that the ILBM counteracts this gravity torque, the ILBM activity.

Table 5.1. Some data from the 11 subjects

subject	male/female	age in years	length in meters	weight in kilograms
A	F	19	1.67	59
B	F	29	1.74	60
C	M	24	1.69	67
D	M	34	1.73	68
E	F	21	1.72	60
F	F	22	1.69	54
G	M	22	1.78	61
H	M	19	1.83	63
I	M	27	1.76	75
J	F	18	1.59	59
K	M	25	1.83	72

5.2 Materials and methods.

Eleven healthy subjects (see table 5.1) with a leg inequality smaller than 10 mm walked on a treadmill at a speed of 4.0 km/h. The treadmill (2.60 x 0.60 meters) had a continuously variable speed control from 0.10-14.10 km/h and was insensitive to the influence of reaction forces of the walking person (Kauer et al., 1985). The treadmill was made of conducting rubber and the subjects wore shoes with contacts indicating heel strike and toe off for both feet.

Pelvic rotations in the frontal and sagittal plane were recorded with a 'pelvis girdle', which was firmly strapped onto the pelvis (Vink, 1986b, Van Leeuwen et al., 1988). The 'pelvis girdle' was connected to a potentiometer, which recorded its deviations from a horizontal position (see fig.5.1) with a

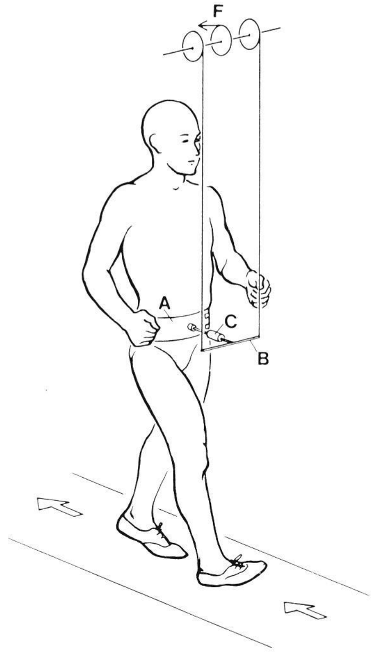


Fig.5.1. The recording of pelvic rotation in the frontal plane. The girdle (A), has been firmly strapped onto the pelvis. The pelvis girdle is fixed to a potentiometer (C) from which the axis is connected to a horizontal bar (B). Both wires attached to the horizontal bar (B) have equal but variable lengths. The wires run over two wheels, which are both connected to a third wheel, which delivers a more or less constant force compensating for the weight of the horizontal bar.

gauged accuracy better than 0.2 degrees (Vink, 1986b) and a frequency response up to 2.5 Hz. Myoelectric activity was recorded using gold-plated home-made bipolar surface electrodes (diameter of 7.5 mm, bipolar center to center distance of 21.5 mm).

The electrodes were applied to the three large parts of the ILBM (according to Jonsson, 1970) as in previous studies (Andersson et al., 1974; Vink et al., 1987 ;see fig. 4.2):

no 1 and 7 (30 mm lateral to L1) recording activity of the multifidus, the spinalis and the longissimus muscle

no 2 and 8 (60 mm lateral to L1) mainly on the longissimus.

no 3 and 9 (10 mm medial to the lateral border of the erector spinae muscle, appr 80-90 mm lateral to L1) mainly on the iliocostalis.

no 4 and 10 (30 mm lateral to L3) mainly on the multifidus

no 5 and 11 (60 mm lateral to L3) mainly on the longissimus

Electrodenumbers 6 and 12 were not applied, because the pelvis girdle partly covered the L5 region.

Due to the complex morphology - i.e. the crossing of the fibers of the longissimus and the iliocostalis - and the interindividual differences, the recordings could contain the combined signals of neighbouring muscles. Therefore the ILBM will be divided into a medial (electrodenumber 1, 4, 7 and 10), an intermediate (no. 2, 5, 8 and 11) and a lateral column (no. 3 and 9) in the functional descriptions.

The simultaneously recorded signals of the foot contacts, the potentiometer and the 10 electrodes have been stored on tape in FM-mode using a 14 channel recorder (Racal Store 14D). The EMG has been filtered (high pass filtered at 30 Hz, 12dB/octave and low pass filtered at 500 Hz, 24 dB/octave) and amplified as much as was possible within the range of the recorder (-1V to +1V), but equally for all 10 EMG channels. Later the recordings have been converted to digital signals with a sample frequency of 1000Hz for further processing on a PDP 11/70 computer. The activity pattern during a stride is computed for each channel by rectifying and averaging the EMG (=RA-EMG) over 48 succeeding strides, a stride being defined as from RHS

(=right heel strike) to RHS. For calculation of the RA-EMG the double support time was divided into eight blocks of equal length, the swing time into 16 blocks (thus, a block in the swing time is different from one in the double support time!). A mean rotation pattern is calculated in the same way but using twice as many blocks.

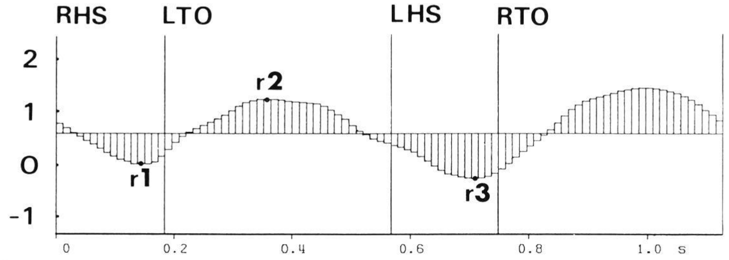
The experiment started with a recording of a subject standing (with extended knees) with the axis of the potentiometer perpendicular to the frontal plane so as to establish the baseline (0 degrees). After 10 minutes of habituation to treadmill walking, 2 minutes recording followed with the axis perpendicular to the frontal plane. This was then repeated with the axis perpendicular to the sagittal plane. From these 22 runs 5 were incomplete due to wire fractures in one shoe. In 2 minutes approximately 100 strides were collected. These samples also included contacts of the swinging leg or switch interruptions. We standardized the test by averaging over 48 strides for all subjects and both planes, as this was the largest number of successive clean strides found in all subjects.

5.3 Results.

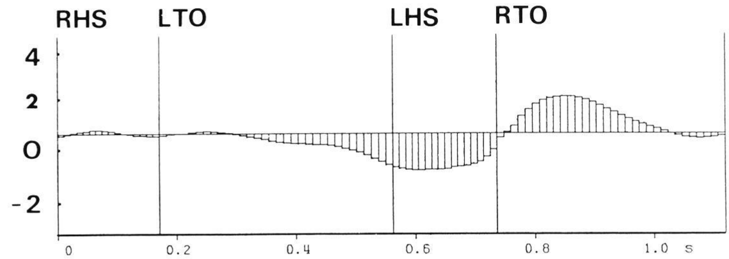
The two runs of one subject resulted in different stride times. During the 48 successive strides the subjects walked with a constant stride time, but after a stop most subjects switched to another stride time showing almost no variation at the same mean speed (4.0 km/h). The difference is significant for 7 out of 11 subjects (two sided t-test for paired samples; $p<0.05$).

The mean durations of phases in 48 strides are summarized in table 5.2. The mean stride time of the eleven subjects (RHS-RHS) is 1.156 sec. The mean values of the two double support times differ significantly for 12 out of 17 complete runs (one sided t-test for paired samples, $p<0.05$; see table 5.2):The RHS-LTO time (mean 0.188s) is longer than the LHS-RTO time (mean 0.182s).

Seven out of eleven subjects have the following characteristics of <u>sagittal</u> rotation (see fig. 5.2a). Between RHS and LTO the ventral part of the pelvis rotates upwards to about level position. At about LTO this rotation is reversed (r1 in fig.5.2a, at 13.5 % of the mean walking cycle) reaching its



A



B

Fig.5.2a and 5.2b. Mean pelvic angles in the sagittal plane of subjects C (3a) and E (3b) during the walking cycle averaged over 48 succeeding strides. Pelvic rotation is given in degrees, a positive angle is a rotation with the dorsal part of the pelvis upwards. The first reverse (r1) occurs at 12.8 % of the walking cycle for subject C, considering RHS-RHS 100 % (averaged over 11 subjects 13.5 %), the second (r2) at 31.3 % (35.2 % averaged) and the third at 63.4 % (60.8 % averaged).

minimum halfway LTO and LHS (r2, at 35.2 %). From this point the rotation is again reversed and the ventral part of the pelvis rotates upwards to about level position (r3, at 60.8 %). The upward movement between LHS and RTO is a repeat of the first half of the walking cycle, which is essentially similar, although its amplitudes may differ significantly.

The reverses of the other four subjects were not detectable and they showed large asymmetries. The largest asymmetry is found for subject E (see fig. 5.2b), almost no rotation in the first part (RHS-LHS) of the walking cycle and 3.1 degrees in the second half (LHS-RHS).

Table 5.2. The relative durations of parts of the walking cycle as a percentage of a stride, averaged over 48 succeeding strides, walking on a treadmill at 4.0 km/h. The mean stride times (RHS-RHS) with standard error of the mean (SEM) are given in seconds. The last column shows the magnitude of pelvis rotations. Due to wire fractures in one shoe subject B, H and J are synchronised on one foot. The mean of 17 complete runs are shown in the bottom row.

subject	RHS-LTO	LTO-LHS	LHS-RTO	RTO-LHS	RHS-RHS	SEM	degr
pelvic rotations in the frontal plane							
A	16.7	33.9	16.1	33.3	1.132s	.019	7.0
B	16.0	34.4	16.0	33.5	1.068s	.014	8.4
C	16.3	34.4	15.6	33.7	1.144s	.016	4.9
D	15.6	34.5	15.3	34.5	1.188s	.019	6.3
E	16.1	34.1	15.7	34.2	1.124s	.013	11.5
F	16.4	34.0	16.1	33.5	1.109s	.020	8.6
G	16.2	33.4	15.9	34.4	1.147s	.020	5.6
H	-	-	-	31.3	1.175s	.028	6.1
I	17.5	33.8	15.3	33.4	1.265s	.036	5.3
J	-	-	-	33.4	1.168s	.030	8.0
K	16.2	33.4	16.0	34.4	1.180s	.012	4.3
pelvic rotations in the sagittal plane							
A	16.6	34.2	16.2	32.9	1.103s	.029	1.9
B	-	-	-	33.2	1.098s	.015	2.3
C	16.3	34.1	16.0	33.4	1.128s	.017	1.8
D	15.4	34.7	15.2	34.5	1.207s	.017	0.4
E	15.4	34.7	15.5	34.4	1.122s	.014	3.1
F	16.1	34.5	15.8	33.5	1.131s	.019	2.4
G	16.1	33.2	16.3	34.5	1.161s	.017	2.7
H	-	-	-	31.2	1.207s	.037	4.1
I	17.3	33.5	15.6	33.7	1.238s	.037	3.4
J	-	-	-	33.4	1.160s	.020	2.1
K	16.3	33.4	16.3	34.1	1.180s	.017	0.4
mean	16.3	34.0	15.8	33.9	1.156s		

The pelvic rotation in the frontal plane found for all 11 subjects is characterized by the following events within a stride (see fig. 5.3). During RHS the pelvis is slowly rotating with the left side upwards. Just after RHS the rotation is reversed (r4 in fig.5.3; at 2.1 % of the mean walking cycle for all subjects), and a fast rotation follows, where the left side of the pelvis drops and stops just after LTO (r5, at 21.5 %). Now, the left side of the pelvis slowly rotates upwards to a stable position halfway between LTO and

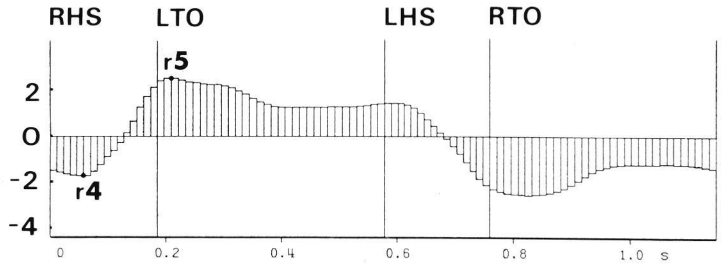


Fig.5.3. Mean pelvic angles in the frontal plane of subject C during the walking cycle (from RHS to the next succeeding RHS) averaged over 48 succeeding strides. Pelvic rotation is given in degrees, a positive angle is a rotation with the right side of the pelvis upwards. The first reverse (r4) occurs at 5.0 % for subject C (averaged over 11 subject 2.1 %), the second reverse (r5) at 18.3 % (21.5 % averaged).

LHS. The pattern of 8 subjects now temporarily shows no rotation, the pelvis maintaining its slightly oblique position for a short period before LHS. Then at about LHS a reversal of the described pattern after RHS is shown. There is asymmetry, six subjects having more deflection with the right side of the pelvis upwards and five with the left side upwards.

ILBM activity started just before or just after heel strike and ended just after toe off for all subjects (see fig. 5.4) as previously described (Dofferhoff & Vink, 1984). In this earlier study the lateral column sometimes showed no activity. This was not seen in the present results. Although large differences in amplitude were found, the lateral column of the ILBM showed activity in both double support phases for all subjects. There was a resemblance between the EMG pattern (see fig. 5.4) and the pelvic rotations in the sagittal plane (see fig. 5.2a), but the correlations between them are low and negative (mean -0.379), which means that a high RA-EMG is found for a position of the pelvis in which the dorsal part is rotated downwards.

As the same amplification is used for all EMG-channels, activity of the left and the right ILBM can be compared so as to estimate its mechanical effect in the frontal plane. In the double support phase the homolateral (with

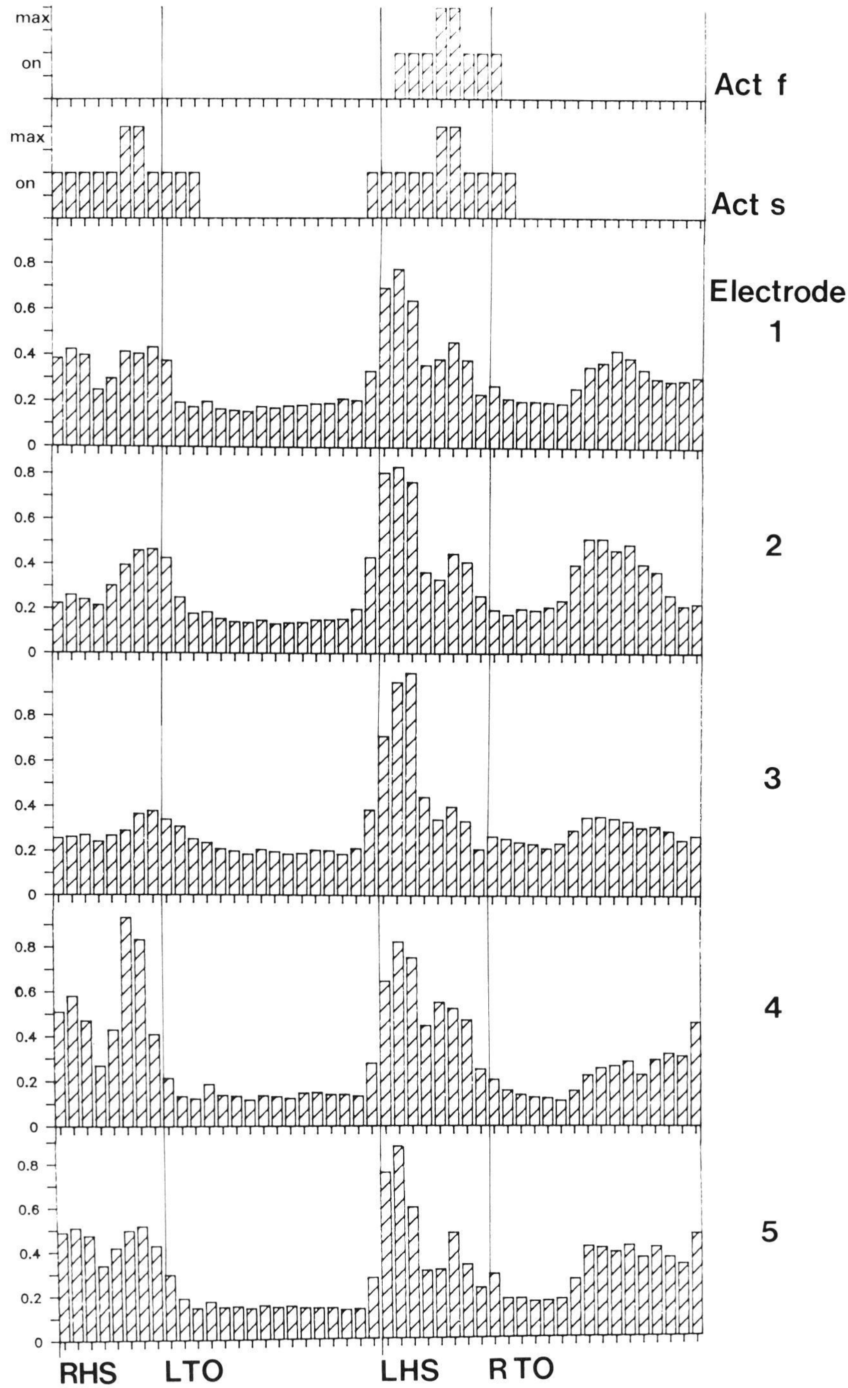


Fig. 5.4. RA-EMG of 5 electrodes positioned on the ILBM of subject C during walking. The electrodenumbers correspond to fig.4.2. Act f and act s at the top of the figure are the predicted activities in the frontal plane, and sagittal plane according to Cappozzo (1983) (derived from his graphics).

respect to toe off) ILBM shows significantly more electromyographic activity than the contralateral for 75.8 % of the electrodes (5 electrodes have been paired (left-right) tested for each of the 8 blocks, for each double support phase and for each run, t-test, $p<0.05$). For electrodes placed on the intermediate and lateral column the asymmetry was higher (86.4 %).

Correlations between RA-EMG and degrees of pelvic rotation in the frontal plane were very low (mean 0.12), but the difference in RA-EMG between the left and right ILBM showed higher correlations with the frontal plane (mean 0.30), especially for the intermediate (mean 0.37) and the lateral (mean 0.41) column.

In the double support phase the RA-EMG of the homolateral (with respect to toe off) ILBM decreased in 75 % of the electrodes (25 % increased). They were rather similar for all electrodes, but the contralateral had a more variable shape (see fig. 5.4).

5.4 Discussion.

Walking patterns have a very individual character, likewise each cycle varies. After a rest 7 out of 11 subjects showed different stride times even at the same speed.

An asymmetry in the walking pattern is shown. The RHS-LTO time is longer than the LHS-RTO time. A phenomenon which also appears in the data of Rosenrot et al. (1980): the mean RHS-LTO time is 0.22 (our data: 0.188) and the mean LHS-RTO time 0.20 (our data: 0.182). This means that the left foot is placed relatively earlier on the ground. This asymmetry may be due to a slight tilt of the treadmill or such unknown factors.

During static extension against resistance the medial parts of the ILBM were found to be recruited earlier than the lateral parts (see chapter four). Such functional differences between the three columns of the ILBM were hard to observe in individual subjects during walking. However, when averaged over all of the 11 subjects, differences with respect to forces in the frontal plane were indeed seen. The intermediate and lateral parts show more asymmetries than the medial parts.

The prediction of Cappozzo (1983) is supported by our results. Indeed the whole ILBM is bilaterally active from about heel strike until one third of the subsequent stance phase (=about toe off of the other foot) to compensate for the moment in the sagittal plane, with a superimposed activity of the homolateral ILBM (related to toe off) to compensate for the moment in the frontal plane.

A discrepancy is found between the moment in the frontal plane as computed by Cappozzo (1983), which showed increase between heel strike and toe off, and our results, in which the RA-EMG decreases in 75 % of cases (see f.i. fig.5.4). This may indicate the existence of a phase lag between electrical and mechanical response, which may indicate a preceding energy build up in the ILBM and surrounding tissue. This storage of strain energy plays an important role in mammalian running (Alexander, 1977).

The negative and low correlations between RA-EMG and pelvic rotations in the sagittal as well as in the frontal plane prove that pelvic rotation is an unimportant factor in determining the moments found by Cappozzo (1983). The pelvic rotations we have found in both planes seem to be valid, because they are similar to those found by Belenky (1973), Inman et al. (1981) and Thurston (1982).

An explanation for the alternating asymmetric activity of the ILBM related to moments in the frontal plane could be found in the laterorotations of the thorax, which mirror the pelvic rotations (Thurston & Harris, 1983). This difference in activity between both ILBM is therefore only indirectly correlated with pelvic rotations. This may be one reason for the relatively low correlation coefficients. Other reasons may be the rather high inter- and intraindividual variation as well as the fact that ILBM activity is actually related to simultaneous movements in three planes.

5.5 Conclusion.

The pelvic rotation does not determine the ILBM activity, probably the changes in the position of the torso are of more importance.

Most of the activity of the ILBM, found in the bipedal phase, is needed to prevent the torso from falling forward. As the moment arms are equal in the sagittal plane, bilateral activity of all parts is found.

Differences in activity during walking between the three columns of the ILBM are small. Only if correlated with movements in the frontal plane were differences found: the intermediate and lateral columns of the ILBM deliver the moments needed in the frontal plane, assuming that the pelvic and thoracic rotations are mirrored. After right heel strike the right ILBM (and after LHS the left) are mostly more active than the left in order to counteract the lateroflexion of the torso. There is a phase difference between the electrical and mechanical activity of the ILBM with respect to forces in the frontal plane.

5.6 Summary.

Gait variables, pelvic rotations in the frontal and sagittal plane and RA-EMG (rectified and averaged EMG) of the three columns of the intrinsic lumbar back muscles (=ILBM) were recorded simultaneously during 48 succeeding strides of 11 subjects on a treadmill.

Bilateral activity is found in all parts of the ILBM during the double support phase. After right heel strike the right ILBM (and after left heel strike the left) show in most cases more activity than the contralateral ILBM. This is especially so in the intermediate and lateral columns, which consist mainly of the longissimus thoracis and the iliocostalis lumborum muscle and less so in the medial column, made up mainly by the multifidus and spinalis muscle. This difference is probably due to the difference in moment arm for the two directions.

Pelvic rotations are described, but no evident relationships between pelvic rotations in the different planes and ILBM-activity could be seen, probably because the changes in the position of the torso are of more importance.

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6. Elastic strain energy in the low back muscles during human walking

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6.1 Introduction

In a previous study (see chapter 5; Vink and Karssemeijer 1988) activity of three columns of the intrinsic lumbar back muscles (=ILBM) was recorded bilaterally with 10 bipolar electrodes. This activity was compared with moments computed by Cappozzo (1983). He estimated forces and moments on the fourth lumbar vertebra during walking. The mass, location of center of gravity, and moments of inertia of body segments were estimated by anthropometrical measurements. The positions and orientations of 5 body segments (head, two arms, torso and pelvis) were recorded during walking with a stereophotogrammetric technique, from which moments in three perpendicular planes were estimated.

Largest were the moments during flexion (moment in the sagittal plane) and lateroflexion (moment in the frontal plane) of the torso. These moments should be counteracted by the ILBM to prevent the torso from falling. The moment in the sagittal plane is counteracted by bilateral activity of the ILBM, and the moment in the frontal plane (lateroflexion) is counteracted by the difference in activity between left and right ILBM. Both moments require an increase of the right ILBM force in the bipedal phase beginning with left heel strike, and an increase of the left ILBM force beginning with right heel strike. So, each force must reach its maximum near toe off.

It is shown in chapter five that the maximum rectified and averaged EMG value (=RA-EMG) of 11 subjects (each walking 96 strides) was found in the bipedal phase. Activity of the right ILBM was found after left heel strike and left ILBM-activity after right heel strike, very similar to the predicted forces. However, in 74% the maximum RA-EMG value was close to heel - strike and this RA-EMG value decreased during the bipedal phase (see fig. 6.1).

Thus, whereas Cappozzo (1983) predicted the maximum force of the right ILBM at the end of the bipedal phase, we found that the RA-EMG reaches its maximum mostly at the beginning of the bipedal phase and its minimum

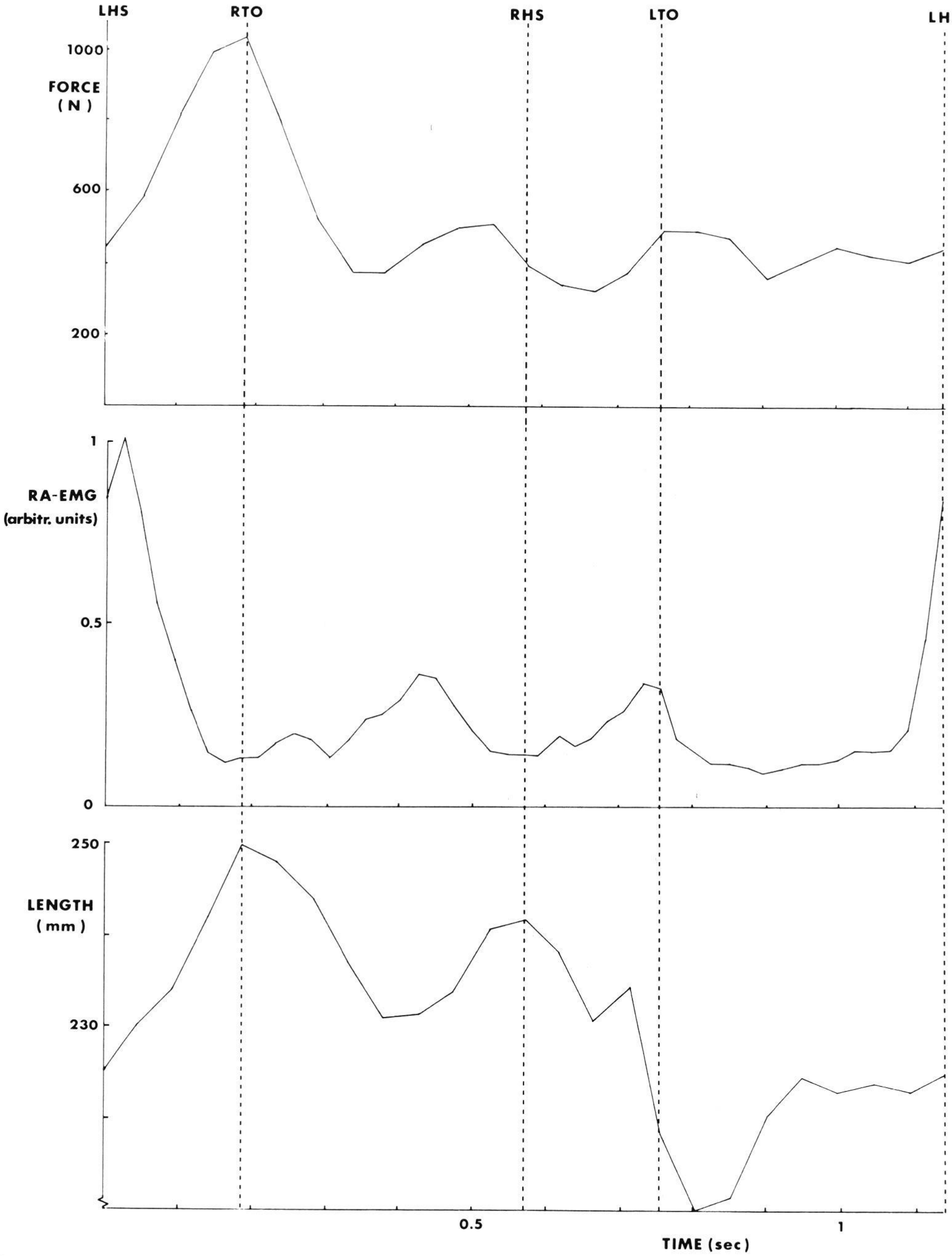

Fig. 6.1 The predicted ***force*** *exerted by the most lateral part of the right intrinsic lumbar back muscles, derived from the results of Cappozzo (1983), the recorded activity in* ***RA-EMG*** *and the* ***length*** *of these muscles calculated with the model (see fig. 2) during a stride. RHS = right heel strike, LTO = left toe off*

at the end. In previous studies (see discussion) it was shown that energy is stored at one stage of the walking cycle and released in another. These considerations led initially to the supposition that during walking elastic strain energy was stored in the left ILBM after right heel strike in the beginning of the bipedal phase and released at the end of the bipedal phase. The same holds for the right ILBM in the bipedal phase after left heel strike.

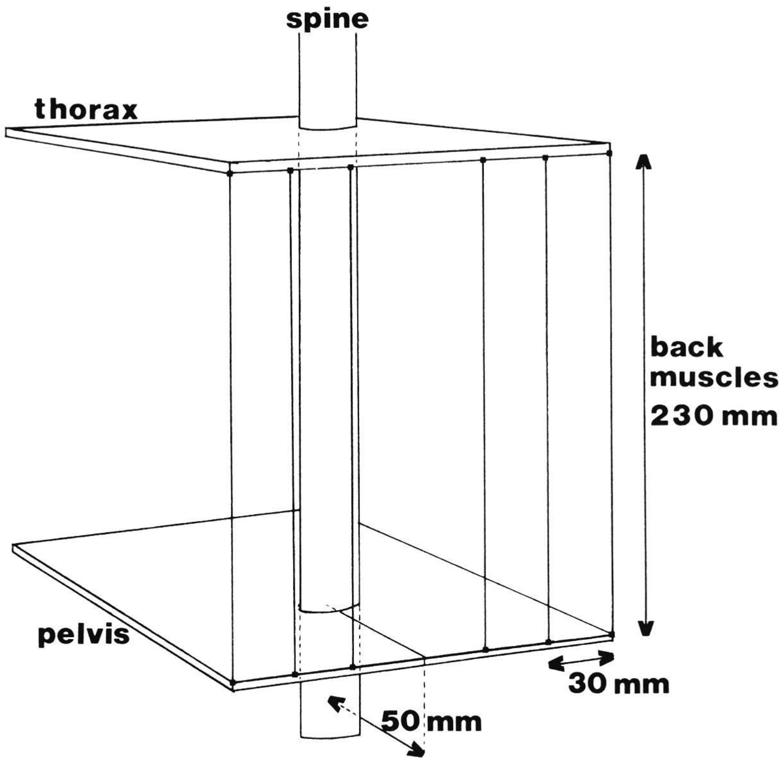

Fig. 6.2 Model of the ILBM to calculate the length of the three columns of the ILBM.

6.2 Materials and methods

To verify this supposition, we developed a simple model of the spine, torso, pelvis and ILBM (see fig. 6.2), and estimated the lengths of the three columns of the ILBM in the different phases of the walking cycle.

In our model, the pelvis and the thorax are represented by transverse plates. In lateroflexion (and also in flexion, extension, and rotation), each plate rotates about one point lying on the plate. These points are located on a vertical line (the simulated spine) with a distance between them of 0.23 m. The three parts of the ILBM are represented by chords between origin and insertion. All origins and insertions are positioned 0.05 m dorsal to the spine. The origins and insertions of the medial column (mainly made up by the multifidus, spinalis and longissimus muscles) are located 0.03 m lateral to the spine, the intermediate column (longissimus muscle) 0.06 m, and the lateral column (iliocostalis lumborum muscle) 0.09 m (see fig. 6.2).

The rotation angles are derived from the data of Thurston and Harris (1983), who recorded the rotations of the pelvis and the thorax in three planes (The pelvic rotations were very similar to the rotations recorded in the EMG experiment of chapter five). These angles were inserted into a transformation matrix of the three-axis Eulerian angle system (Chao 1980). The computed rotations indicate new coordinates of origins and insertions, and the length between origin and insertion can be calculated using the Pythagoras' algorithm.

6.3 Results

The length of the right ILBM is maximal at right toe off (=RTO), exactly the stage were the maximum force is needed (see fig. 6.1). The lengthening is especially large for the lateral column. The same is found for the left ILBM at LTO.

At left heel strike (LHS), the right ILBM is slightly shortened, while the rate of lengthening is positive. The rate of lengthening remains positive during the subsequent part of the bipedal phase (and the muscle becomes eccentrically active immediately after LHS), while the RA-EMG decreases during this phase reaching almost zero at right toe off. As the RA-EMG is an indication of the degree of muscle excitation, an increase in force normally corresponds with an increase in RA-EMG. However, the degree of muscle excitation required to produce a given force is smaller, when the active muscle is forcibly stretched (Bigland and Lippold, 1954), which is the case in our situation. This does not fully explain the observed phenomenon: since the rate of lengthening is constant and the force increases, an increase in excitation (RA-EMG) would be expected, whereas the opposite, a decrease in excitation, is found (see fig. 6.1).

Therefore, the maximum force needed according to Cappozzo (1983) must have been produced passively by the lengthening of the muscle.

6.4 Discussion

Muscles and tendons have elastic properties. Energy is saved by storing elastic strain energy at one stage of the walking cycle and releasing it in another. Alexander and Bennet-Clark (1977) showed for instance that the elastic properties are exploited during human running as the plantar flexors of the ankle first stretch after foot contact and then shorten at toe off. The same principle was found in hind limb movements of the dog (Wentink 1977): Energy, which is stored at the beginning of the support phase, is released at the end of the support phase and contributes to propulsion.

Assuming that the data of Cappozzo (1983), Thurston and Harris (1983), and chapter five are comparable, this study shows that elastic strain energy is used during human walking in the lumbar back. However, the supposition that energy is stored and released in the same stage was wrong. The passive force produced by the lengthening of the ILBM and their surrounding tissue in the bipedal phase is needed in that same stage. The contractile properties play only a minor role in producing the maximum force, even if the delay between electrical activity and mechanical effect is considered. This latency period is approximately 50 ms (Sherif et al. 1983), whereas the time difference between the maximum in the recorded signal and the maximum force is about 130 ms:

Thus, human walking is efficient. At the stage of the walking cycle in which most force is needed to prevent the torso from falling forwards and sidewards (Dofferhoff & Vink, 1985), the pelvis is rotated in such a way that the force can be delivered passively.

Fick (1911) and Floyd and Silver (1956) showed that the ILBM is not active when the trunk is fully flexed. These investigators suggested that passive forces produced by ligaments counteract the gravity moments in full flexion. However, muscle itself also produces passive forces, when it is elongated. The endomysium, the epimysium and perimysium in the muscle are the parallel elastic elements, which are able to store strain energy. Thus, the ILBM and their surrounding tissue (including the ligaments) produce passive forces during full flexion, but as shown in this study also during walking.

6.5 Summary

A simple model of the thorax, pelvis and three columns of the intrinsic lumbar back muscles (=ILBM) was constructed. The model was used to study the length of the ILBM during the different stages of the walking cycle.

The length of the right ILBM (especially the lateral column) was largest at right toe off, exactly the stage of the walking cycle in which most force was needed to prevent the torso from falling forwards and laterally.

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7. Leg length inequality, pelvic tilt and lumbar back muscle activity during standing.

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accepted in Clinical Biomechanics

7.1 Introduction

Some clinicians state that leg length inequality causes pelvic tilt and consequently low back pain (Helliwell, 1985), whereas others cast doubt on the existence of an association between leg length inequality and back pain (Grundy & Roberts, 1984). These conflicting opinions stress the need for studies that verify effects of leg length inequality on the mechanics of the back.

Taking the view that leg length inequality causes back pain, the question arises: "which inequality should be corrected?". Nicholls (1960) considered a difference of 12.5 mm to be important, while Helliwell (1985) corrected at 20 mm. A careful decision should be made, because a shoe raise or surgical correction burdens the patients in financial and physical respect.

Whenever there is a lateral pelvic tilt due to a leg length inequality there exists a functional scoliosis of the lumbar spine to keep the torso upright (Rush & Steiner, 1946). It is unknown to which extent activity of the ILBM is needed for stabilising the trunk.

We developed a method to observe pelvic rotations (Vink, 1986; Van Leeuwen et al., 1988) and EMG of different parts of the ILBM to study the stabilising function of these muscles (Vink et al., 1987). With these methods we investigated the influence of an artificial leg length discrepancy (=ALLD) on pelvic tilt and activity of the ILBM.

7.2 Materials and methods.

Ten male subjects (mean age 23.7 years, SD 3.03; mean leg length 955 mm, SD 57) with no history of back disease were studied.

Recordings were made, while the subjects stood upright with extended knees and 50 mm distance between both feet. An ALLD was created by putting different boards of 5.0 mm under the right foot. In 5 subjects (a to

e) the experiment started with an increase, in the other 5 subjects (f to j) the leg length discrepancy was diminished.

The pelvic tilt was recorded with a technique, which has been described extensively elsewhere (Vink, 1986; Van Leeuwen et al., 1988): A girdle is firmly strapped around the pelvis; the girdle is fixed to two potentiometers, which record simultaneously angles of the bony pelvis around a frontal and a sagittal axis. Dorso-ventral radiographs made of a fat subject wearing the girdle with different artificial leg length discrepancies showed a maximal shift of the girdle of 0.09 mm. The maximal difference found in repeated experiments was 0.06 mm, which means that its accuracy is better then 0.15 degrees. The subjects were instructed not to rotate around the frontal or longitudinal axis.

Surface EMG was recorded by a procedure described in detail in chapter five.

The EMG recordings started when the EMG was stabilised (approximately after 30 seconds standing with the ALLD). The digitized EMG was rectified and averaged (RA-EMG) over 5 successive samples of 1 second. The sample with the maximum RA-EMG was chosen and calculated as a percentage of the RA-EMG during a maximal extension against resistance (see chapter four). This procedure was chosen because of its relative high reproducability (always higher then 76%), which is especially so for the larger leg length discrepancies.

The RA-EMG value (y-value) was plotted as a function of the ALLD (x-value). The x-value, for which the difference in y value between the fitted line and the recorded RA-EMG value was largest, was calculated. This indicated for which ALLD the RA-EMG increased.

7.3 Results.

The lateral pelvic tilt increased linearly with the ALLD (see fig. 7.1; mean 0.216 degrees/mm; corr. 0.9907; $p<0.002$). In four subjects (b, d, g and i) the maximum lateral pelvis tilt was less then 11 degrees. Their maximum tilt occurred at about 50 mm ALLD. The other six subjects could reach lateral pelvic tilts at 55 mm ALLD. The largest tilt was 11.9 degrees.

Table 7.1. The ALLD (in mm) for which the difference between a linear fitted line and the recorded RA-EMG value was largest, indicating the point of the RA-EMG increase. - means that there was no increase in RA-EMG. LLI is the natural leg length inequality, + is longer right leg (difference in distance from the anterior superior iliac spine to medial malleolus).

		electrodenumber				
subject	LLI (mm)	1	2	3	4	5
a	-5	50	-	25	45	45
b	0	35	25	35	35	35
c	10	35	35	35	30	35
d	0	35	35	35	35	30
e	15	20	25	25	15	20
f	0	45	30	40	45	45
g	0	30	30	30	30	30
h	0	40	40	40	40	40
i	5	40	35	30	40	40
j	-5	-	-	-	35	35
	mean	37	32	33	35	36

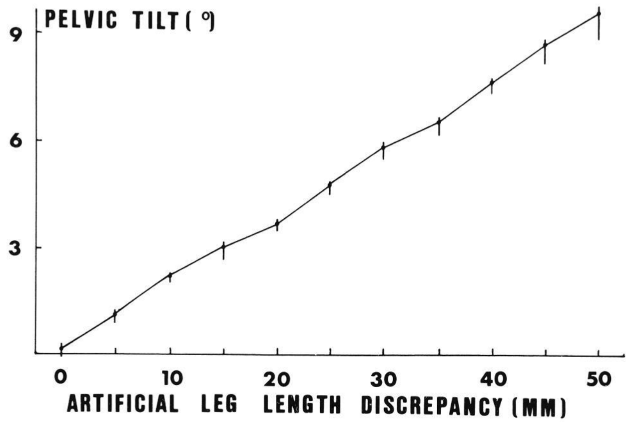

Fig. 7.1. The increase in pelvic tilt in degrees for different ALLD (=artificial leg length discrepancy). The vertical line around each point is the standard deviation.

For all subjects the RA-EMG increased only unilaterally and was always at the longer leg side. For small leg length discrepancies the RA-EMG was level or slowly increasing. Above a certain leg length discrepancy (mean 34.4 mm) the RA-EMG increased more strongly (see fig. 7.2). The point from which the RA-EMG increased, varied for the subjects and electrode sites (from 15 to 50 mm; see table 7.1). The medial electrodes showed most increase related to their maximum RA-EMG (up to 2.2 % of the maximum RA-EMG), whereas the lateral electrodes showed the least increase (up to 0.7 % of the maximum RA-EMG).

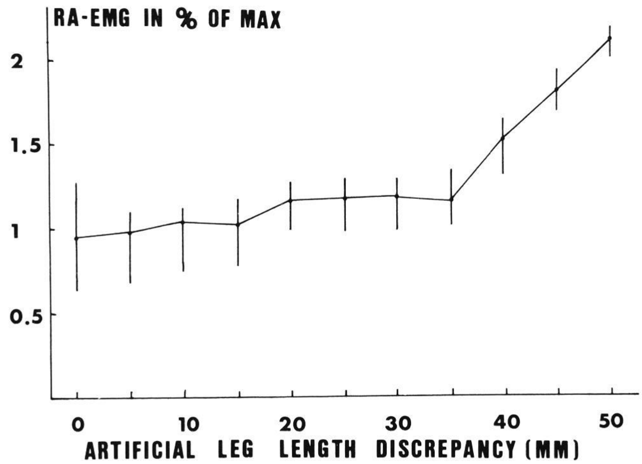

Fig. 7.2. The increase in RA-EMG for different ALLDs for an electrode located 30 mm right from the spinal process of L3 averaged over 10 subjects. The vertical line around each point is the standard deviation.

7.4 Discussion

This study confirms that a leg length inequality causes pelvic tilt while standing with extended knees. It could be observed that a functional scoliosis of the lumbar spine compensated for the tilt of the pelvis. As a result the ligaments, connective tissues and muscles are lengthened at the short leg side. Thus a moment is exerted, which has to be compensated for at the contralateral side by activity of the intrinsic lumbar back muscles in order to keep the torso upright.

Force produced by lengthening of passive structures in the trunk was also postulated by Thorstensson et al. (1985). At the end of a slow and large lateroflexion they found activity of ipsilateral muscles to counteract the passive force at the contralateral side.

In a previous study (Vink et al., 1987) RA-EMG of the medial column of the ILBM increased linearly with increasing forces. The subjects extended the back against resistance up to the maximum force in ten steps under isometric conditions. In the present study RA-EMG of the ILBM increases non-linear with ALLD, probably to compensate a non-linear increase of the passive force at the longer leg side. The reason for this non-linearity could be:

a. because the lengthening of the tissues is not only influenced by the lateral pelvic tilt but also by curvatures of the spine in the transversal and sagittal plane to correct the pelvic tilt; Pearcy (1980) showed that rotations in the frontal plane of the lumbar spine are accompanied by small rotations of the vertebrae in the transversal and sagittal plane.

b. because the force produced by the passive lengthening is only substantial above 35 mm.

In view of the increase in RA-EMG activity with the more substantially inequalities one could conclude that at least clinical corrections are necessary for artificial leg length discrepancies of 35 mm or more, because the additional activity of the ILBM would increase the loading of the spine. This does not necessarily mean that small leg length inequalities (<35 mm) may be harmless. Firstly, for some subjects the ILBM activity increases at 15 mm (e.g. subject e), secondly, the loading or deformation of the spine could also increase without ILBM-activity (for instance by passive stretching of tissues) and thirdly, long term effects of a natural leg length inequality still need to be investigated. On the other hand it is possible that a structural scoliosis in a clinical leg length inequality is in an opposite direction from that expected. In these cases a correction may increase loading on the spine and thus is inappropriate.

Thus, the results presented are only indicative of the level of inequality to be corrected. Studies with patients with a natural leg length discrepancy are required to apply the value of these findings.

7.5 Summary

The influence of an ALLD on lateral pelvic tilt and on activity of the ILBM was investigated. An ALLD of up to 50 mm was created by putting boards of different height under the right foot.

Lateral pelvic tilt increased linearly with increasing artificial leg length discrepancies. The rectified and averaged EMG of the ILBM showed a small increase at the longer leg side. It increased non-linearly with an increment in slope above a certain ALLD (mean 34 mm).

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8. Lumbar back muscle activity during walking with a leg inequality.

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Acta Morphologica Neerlando-Scandinavica 25:261-271, 1987

8.1. Introduction

There is still considerable debate about the hypothesis that leg length inequality (=LLI) causes pelvic tilt and consequently chronic low back pain. Among others, Helliwell (1985) claimed good effects of a shoe raise of the shorter limb for patients with back pain, whereas, for instance, Grundy and Roberts (1984) found that the percentage of subjects with a LLI in back pain patients differed not significantly from the percentage of normal subjects.

It should be realized that a pelvic tilt as an effect of LLI is only established while standing. Whether this effect should also apply to walking is still unknown. Additionally, the influence of a LLI on back muscle activity is unknown. For this reason we decided to study the influence of an artificial leg length discrepancy (=ALLD) on rotations of the pelvis, on back muscle activity and on stride times while walking.

8.1.1. The functions of different columns of the ILBM during walking

Many authors describe the function of the whole ILBM and not for the individual parts of the muscle (Andersson et al., 1980; Battey & Joseph, 1966; Brauer et al.,1986; Carlsöö, 1964; Marras et al., 1984; Nemeth & Olsen, 1986; Thorstensson et al., 1985). Considering the complex structure of the ILBM (Bogduk, 1980; Bustami, 1986), a functional differentiation between its components should be ideally aimed at. In this study the ILBM is subdivided into three columns (see fig. 4.2). For technical reasons a subdivision into more than three columns is as yet unfeasable (Vink et al., 1988).

In earlier studies two periods of ILBM activity have been found during a stride (Dofferhoff & Vink, 1985; Vink & Karssemeier, 1988). The following mechanical basis for this activity is postulated:

Bilateral activity around heel strike is needed to counteract the trunk flexion which is caused by the decelerating pelvis. The superimposed unilateral activity around contralateral heel strike is needed to counteract the lateral bending of the spine. Especially the lateral column and to a lesser extent the intermediate column counteract this lateral bending. Around homolateral toe off the lateral column shows no activity but probably produces its force by passive lengthening (Vink & Karssemeijer, 1988).

8.1.2. Predicted activity of the ILBM during walking with ALLD

It is to be expected that compared to walking with equal leg lengths, bilateral activity of the ILBM increases at heel strike with the raised limb, because of an increased deceleration during heel strike of the raised limb.

If a LLI causes pelvic tilt during walking as in standing, another prediction can be made. The homolateral activity of the ILBM should decrease during toe off of the shorter limb, because an ALLD causes pelvic tilt and lengthens the passive elements at the side of the shorter limb.

8.2. Materials and methods,

20 subjects (see table 8.1) with no history of back disease walked on a treadmill with different ALLDs at a speed of 4.0 km/h. The influence of an ALLD on stride times, on pelvic rotation angle and on ILBM-activity was recorded.

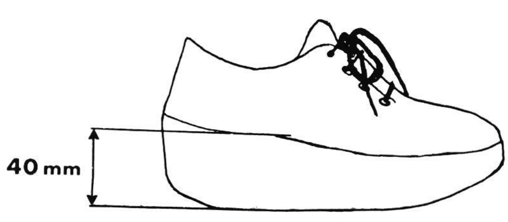

fig. 8.1. The form of the raised shoe (40 mm) used in this study.

The treadmill was made of conducting rubber and the subjects wore shoes with contacts indicating heel strike and toe off for the left as well as for the right foot. The raised shoes (10, 20, 30 and 40 mm) were made with a sole which allowed the foot to roll off (see fig. 8.1). The treadmill (2.60 x 0.60 meters) had a continuously variable speed control from 0.10-14.10 km/h and was insensitive to the influence of reaction forces of the walking person (Kauer et al., 1985).

The rotations of the pelvis in the frontal and sagittal plane were recorded simultaneously with a 'pelvis girdle', which was firmly strapped onto the pelvis (Vink, 1986; Van Leeuwen et al., 1988). The 'pelvis girdle' was connected to two potentiometers, which recorded deviations from a horizontal position, which had a gauged accuracy better than 0.2 degrees and a frequency response up to 2.5 Hz. The mean pelvic rotation angle during walking was calculated by establishing the maximum and minimum pelvic angle for each stride and averaging them over 10 strides.

Myoelectric activity was recorded by a procedure described in detail in chapter 5.

The signals indicating left and right foot contact, the signals of both potentiometers and the 10 EMG signals of the ILBM were stored on tape in FM-mode using a 14 channel recorder (Racal Store 14D). The EMG was filtered (high pass at 30 Hz, 12 dB/octave and low pass at 500 Hz, 24 dB/octave) and amplified as much as possible within the range of the recorder (+/- 1V), but equally for all 10 EMG channels. Later all the recordings were digitized with a sample frequency of 1000 Hz for further processing on a PDP11/70 computer. The activity pattern was computed for each channel by rectifying and averaging the EMG (=RA-EMG) over 48 succeeding strides. For calculation of the RA-EMG the double support time was divided into eight blocks of equal length, the swing time into 16 blocks. A mean rotation pattern was calculated in the same way but using twice as many blocks.

The experiments started with a recording of a subject standing (with extended knees) with the axis of the potentiometers perpendicular to the sagittal and frontal plane to establish the baseline (0 degrees). After 10 minutes of habituation to treadmill walking, 10 strides were recorded. Then, for 10 subjects (A, D, E, G, K, L, N, O, S and T) the right shoe was raised 10, 20, 30 and 40 mm respectively. After recording the new baseline with an ALLD (with extended knees), a habituation to walking with an ALLD was performed during a period of one minute, and again 10 strides were recorded. Then the whole procedure was repeated for a left shoe raise of 0, 10, 20, 30 and 40 mm respectively. The experiment ended with a recording of the baseline and normal walking. The other 10 subjects were recorded likewise, but in a reverse order (40, 30, 20, 10, 0 mm ALLD).

Differences in RA-EMG between walking with and without an ALLD

Table 8.1. Data of the 20 subjects used in this experiment. The influence of an ALLD on the pelvic rotation angle in the frontal plane is shown in the last two columns.

subject (male/ female)		length (mm)	weight (kg)	leg length (mm)	LLI (mm)	pelvic tilt standing (degr./mm)	pelvic tilt walking (degr./mm)
A	m	1695	70	886	0	0.134	0.012
B	f	1765	64	930	-6	0.141	0.024
C	m	1735	72.5	924	-2	0.141	0.026
D	m	1825	66	950	10	0.147	0.007
E	f	1595	58.5	820	10	0.147	0.022
F	f	1760	63	892	2	0.157	0.020
G	m	1830	66	966	0	0.160	0.033
H	m	1785	68.5	942	8	0.160	0.037
I	f	1710	67.5	906	-6	0.166	0.042
J	f	1695	66	882	6	0.168	0.035
K	m	1830	69	982	12	0.179	0.030
L	m	1850	80	1034	2	0.180	0.032
M	f	1715	60.5	898	-4	0.192	0.036
N	f	1740	58.5	906	0	0.194	0.035
O	m	1850	68.5	994	0	0.196	0.050
P	f	1730	67	865	2	0.196	0.051
Q	m	1965	86.5	1072	4	0.197	0.060
R	f	1850	83.5	990	-2	0.200	0.059
S	m	1865	77.5	964	-6	0.211	0.070
T	f	1805	69	960	2	0.214	0.069

were tested for each subject (one sided t-test for paired samples; $p<0.05$). Walking without an ALLD was defined as the average of the three runs (one at the start, one in the middle and one at the end of the experiment).

8.3. Results

8.3.1. ALLD and stride times

Walking with an ALLD produced an increase in the duration of the swing phase (see fig. 8.2) and a decrease in the duration of the stance phase of both limbs with respect to normal walking. The difference was significant for 20 mm or more ALLD, except for the influence of a left raise on the right swing phase, which was only significant at 40mm ALLD. The duration of the stride was not changed significantly (mean duration = 1.186 s).

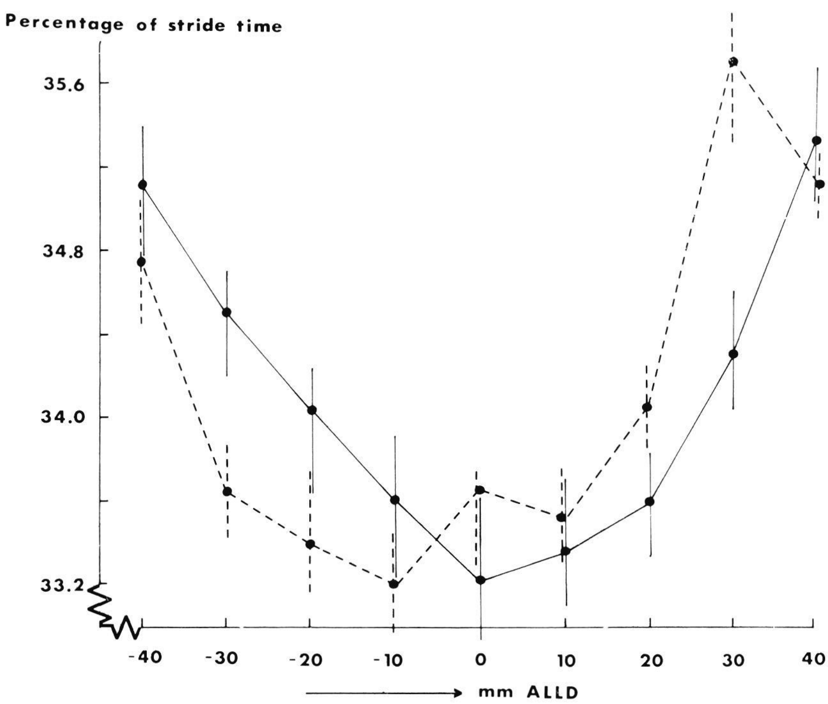

fig. 8.2. The influence of an ALLD on the swing phase time (walking 4.0 km/h) averaged over 20 subjects. The duration is presented in % of the RHS-RHS time;. + is a raise to the right foot; - is a raise to the left foot; the continuous line is the swing phase time of the left leg, the dotted of the right. The vertical line around each point is the standard deviation.

8.3.2. ALLD and pelvic rotation

Pelvic rotations in the sagittal plane were influenced by an ALLD, but the differences were small (about 0.3 degrees, too close to the accuracy of the system, also because high frequencies, up to 5 Hz, were involved (see Van Leeuwen et al., 1988)).

Walking with an ALLD produced a mean pelvic rotation angle (in the frontal plane), which increased linearly with a raise to the homolateral side (38 degrees/m; see fig. 8.3). Related to the increase in standing with stretched knees (174 degrees/m), this raise was small. For one subject (D) the influence of an ALLD on pelvic rotation angle during walking was too small to be measured. The largest pelvic rotation angle during walking was found in subject S (70 degrees/m)(see table 8.1). Those subjects who had a large pelvic rotation angle during walking also showed the largest rotation angle while standing. A correlation of 0.878 was found between them.

Most subjects walked with a slightly different baseline in the

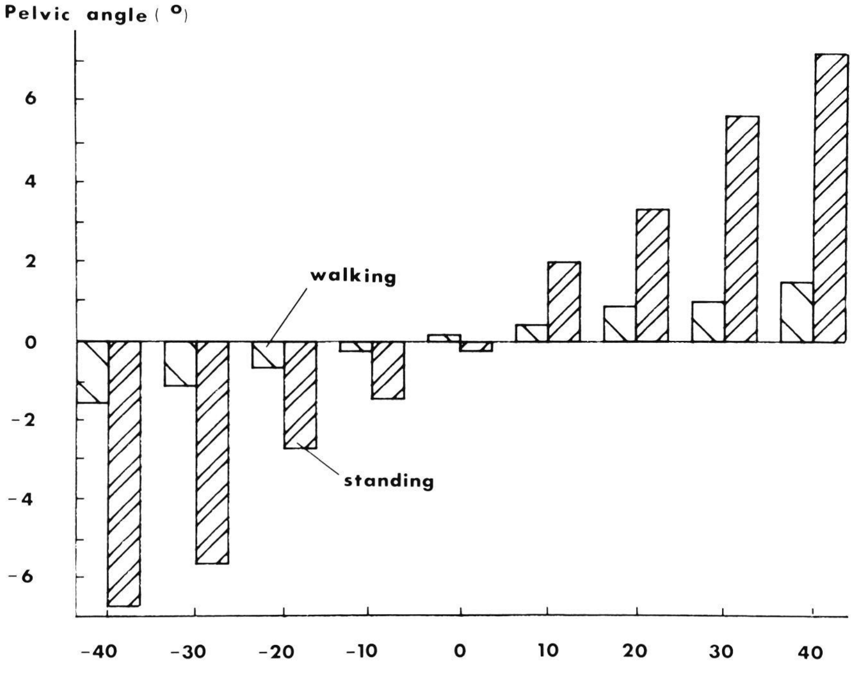

fig. 8.3. The influence of an ALLD on degrees of pelvic rotation angle in the frontal plane during walking (4.0 km/h) and while standing averaged over 20 subjects; + is a raise to the right foot; - is a raise to the left foot.

frontal plane than was established while standing. For three subjects (D, E and K), who had a LLI of 10 mm or more due to a longer right limb, the baseline during walking was clearly different from standing (mean 1.7 degrees with the left side of the pelvis upwards during walking).

8.3.3. ALLD and EMG of the ILBM

For each of the 48 blocks in a stride, EMG-amplitudes with and without an ALLD were compared. In the bipedal phase after heel strike of the raised limb, only in 5.3% of the electrodes significant differences are found. However, in 4.8% of the electrodes the three runs without an ALLD differed also significantly. Even at 40 mm ALLD not more than 4 out of 20 subjects (subject I,Q,S and T) showed significant increases in a part of the stride (in

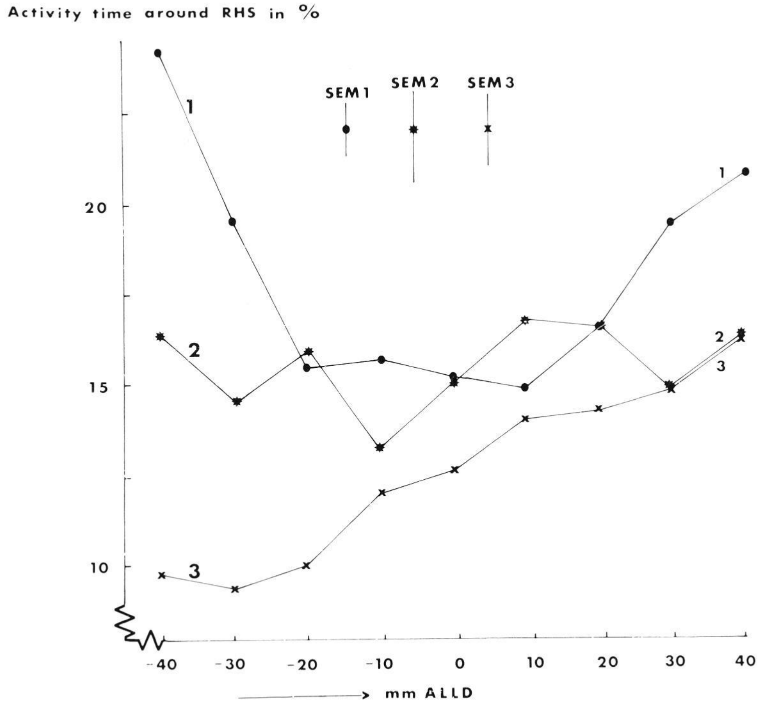

fig. 8.4a. The influence of an ALLD on the activity time of three columns of the ILBM around left heel strike during walking at a speed of 4.0 km/h. Curve 1 is of electrodenumber 1 (see fig. 5.2). SEM1 is standard error of the mean of electrodenumber 1. The activity time is presented in % of the RHS-RHS time averaged over 20 subjects; + is a raise to the right foot; - is a raise to the left foot.

93% of the electrodes on the contralateral side, 2 blocks before and 4 blocks after toe off of the shorter limb).

Most differences were not significant due to the relatively small increase in mean amplitude of the RA-EMG together with a large increase in standard deviation of this RA-EMG.

The activity time of the ILBM (an amplitude above 20% of the maximum RA-EMG during walking is considered as activity) was more clearly influenced by an ALLD:

- the prediction that ILBM-activity increases around heel strike of the raised

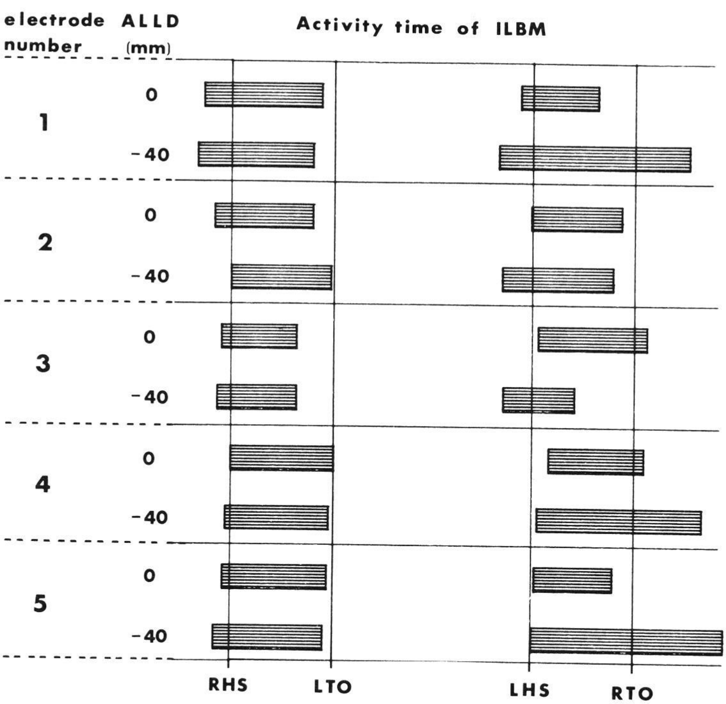

fig. 8.4b. The activity time of the ILBM walking without an ALLD and walking with a raise to the left limb of 40 mm averaged over 20 subjects. The electrodenumbers correspond to fig. 5.2.

limb is supported. The activity time increases significantly for 83.5% of the electrodes around heel strike if the limb is raised 30 mm or more (see fig. 8.4a), especially for electrodenumbers 1, 4, 5, 7, 10 and 11 (93%). Activity started equally for electrodenumbers 4, 5, 10 and 11 with and without an ALLD, but significantly earlier for 92% of electrodenumber 1, 2, 3, 7, 8 and 9 (see fig. 8.4a and b) with an ALLD. No significant differences were found around heel strike of the shorter limb (see fig. 8.4b).

- the prediction that homolateral activity decreases around toe off of the shorter limb is partly supported. The activity time decreases significantly in 97% for electrodenumbers 3 and 9 at 30 mm or more ALLD, but is unchanged for electrodenumber 2 (see fig. 8.4). For electrodenumbers 1, 4, 5 (at RTO), 7, 10 and 11 (at LTO) a contralateral increase is even found (63%).

8.4. Discussion

8.4.1. Stride times and pelvic rotations

The stride times and pelvic rotations are discussed firstly as they interfered with the predicted ILBM-activity.

Rotations in the transversal plane have not been recorded. The shown increase of the swing phase time together with the decrease of the stance phase time for both limbs, could imply more transversal rotations of the pelvis, perhaps to such a degree that their effect on ILBM activity is not negligible.

Influences of an ALLD on pelvic rotations in the sagittal plane were too small to measure.

The fact that the pelvic rotation angle in the frontal plane changes considerably less during walking than while standing with an ALLD (see fig. 8.3), shows that during walking an ALLD is largely compensated below the pelvis. Thus, the assumption that an ALLD causes pelvic tilt during walking is only true for a very small part and it is impossible to verify the second prediction.

8.4.2. Pelvic rotation and ILBM activity

In spite of the small influence of an ALLD on the pelvic rotations the activity time of the ILBM is clearly influenced. This supports conclusions of an earlier study in this journal (Vink & Karssemeijer, 1988), that activity of the ILBM is more related to thoracic movements than to pelvic rotations during walking. The relationship between pelvic rotation and ILBM activity is only shown indirectly for large pelvic angles.

At toe off of the shorter limb the force, produced by the passive lengthening of the homolateral ILBM, is enlarged. As a result the torso would bent to the homolateral side and contralateral ILBM-activity is needed to prevent the torso from falling sidewards. This is supported by the fact that the four subjects with a relative large pelvic rotation angle (I, Q, S and T) showed a significant increase in RA-EMG of the contralateral ILBM at toe off of the shorter limb with a 40 mm ALLD compared with normal walking.

8.4.3 Predicted ILBM activity

The predicted changes in ILBM activity were too small to be seen in amplitudes of the RA-EMG due to the large variation in RA-EMG during walking with and without an ALLD. Differences were only observed in duration when 10 steps are averaged.

The bilateral activity of the ILBM around heel strike is explained previously (Dofferhoff & Vink, 1985) by the need to counteract the moment of inertia of the upper part of the trunk, produced by the decelerating effect of the heel strike. Raising one foot will increase the decelerating effect and, indeed, the activity time of the ILBM increases around heel strike of the raised limb.

A decrease of activity time for the homolateral ILBM (with respect to toe off of the shorter limb) was only found in the lateral column of the ILBM, probably because in this column muscle lengthening is largest. The increase in the medial and intermediate column could be due to the superimposed force needed in the sagittal plane. A thought supported by the result that the activity time is shifted towards heel strike of the raised limb with respect to normal walking.

8.5. Conclusion

The hypothesis that leg length inequality causes pelvic tilt applies to standing. However, during walking the influence of an ALLD on pelvic tilt is small compared with standing. Thus, no support is found for the hypothesis that walking with an ALLD produces low back pain, caused by pelvic tilt.

Nevertheless, changes were seen in the duration of the ILBM activity, which supports the hypothesis of a previous study (Vink & Karssemeijer, 1988) that the pelvic rotation does not determine the ILBM activity. Changes in the position of the torso are of more importance.

8.6 Summary.

The influence of an artificial leg length discrepancy (=ALLD) on stride times, pelvic rotations and activity of the intrinsic lumbar back muscles (=ILBM) was investigated for 20 subjects. An ALLD was created by shoes with a raised sole.

Walking with an ALLD produced an increase of the swing phase time and a decrease of the stance phase time for both feet. The influence of an ALLD on pelvic rotations in the sagittal and frontal plane and on ILBM-activity was small.

Changes in pelvic rotations in the sagittal plane were too small to observe. The mean pelvic rotation angle in the frontal plane was changed 1.52 degrees when walking with an ALLD of 40 mm (6.9 degrees while standing with an ALLD of 40 mm with extended knees).

Only small changes were found in activity time due to an ALLD (not in EMG-amplitude). The activity time of the ILBM around heel strike of the raised limb was increased and unilaterally shifted from toe off in the direction of heel strike with the raised limb.

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9. Summary

An impressive amount of research in the field of back pain is mainly focussed on curative, biomechanical, ergonomical, aetiological or epidemiological aspects. However, the cause of back pain is still unknown. It is also unknown how the back muscles should be used to avoid back pain.

In this thesis methods have been developed which allow the study of back muscle activity in more detail. These methods could be applied in an analysis of optimal recruitment patterns within the scope of back pain prevention. Such patterns could be studied in back pain free subjects who have performed hazardous activities over years. The established patterns of an optimal muscle activity may then be taught to high risk groups while environmental conditions to enable such optimal use should be created (e.g. diminishing loads).

The aim of the research presented in this thesis was primarily to study the way lower back muscles can stabilise the spine under various loading conditions. Parts of these conditions may cause back pain in the long run, such as exerting excessive muscle forces or standing and walking with artificial leg length discrepancies up to 40 mm.

According to the anatomy three larger parts of the back muscles can be distinguished in the lumbar region (see fig. 1.2): medially the multifidus muscle, intermediately the longissimus muscle and laterally the iliocostalis muscle. In this thesis the coordinated activity is object of study. Together these three will be referred to as the ILBM (=intrinsic lumbar back muscles). The **second** chapter deals with some recent contributions to the anatomical descriptions of the ILBM.

Given this anatomical subdivision, together with the results of a pilot study showing a difference in function between the multifidus muscle and the iliocostalis muscle during walking (Dofferhoff & Vink, 1985), a further study of the functional differences of these three muscle parts was undertaken.

In the **third** chapter it was tested whether a specific activity of the three columns could be recorded with the electrode location of Andersson et al. (1974)(see fig. 4.2). It is shown that a differentiated activity of three columns can be recorded with 12 bipolar surface electrodes. An EMG recording with more electrodes at shorter distances to each other is not worthwhile with the present method, because the level of cross-talk will increase unacceptably. This means that too much activity is recorded from muscles (or parts of muscles) other than for which the electrodes are meant.

It was shown that the correlation between two raw EMGs increased, when the distance between their electrodes decreased. When one of the two correlated EMGs was shifted in time a maximum was found. These maxima in the cross-correlation coefficient function (CCCF) were 0.51, 0.22 and 0.15 for electrodes located 30, 60 and 90 mm apart respectively (the absolute averaged over 8 subjects). Other factors such as static versus dynamic experimental conditions, the level of activity, the ECG, noise, power-line-induced-a.c.-components and resistance of biological material have less influence than the distance between electrodes, but nevertheless they do influence the CCCF. This means that these factors should be controlled in comparing two EMGs.

In the **fourth** chapter EMG is recorded under isometric conditions during force exertion varying from 10 to 100 % of the maximal voluntary contraction (=MVC). The subjects stood in the upright position exerting a force during an extension against resistance.

A functional difference between the three columns was found (see fig 4.3): the RA-EMG (=the rectified and averaged EMG of one second) of the medial column increased close to linear with an increasing external force, the intermediate column increases its EMG mainly near maximum forces and the lateral column increases its EMG even later. No significant differences were found between the different levels within a column. Fatigue and the sequence in force levels (increasing or decreasing in time) had little influence on the curve of the force-EMG relationship.

The three columns seem to behave similarly when low EMG-values were compared. Therefore, Andersson et al. (1974) showed no significant differences between the three columns during sitting and standing. However, using a wide variation of force levels, we observed that at submaximal extension against resistance the medial column contributes relatively more than both other columns. At present we can't give an explanation for this phenomenon, probably muscle fibre length and fibre type are of importance.

In the **fifth** chapter functional differences between the three columns were shown during walking. During heel strike the three columns behave similarly: bilateral activity was found in all columns to prevent the torso from falling forward (in fact the inertial forces with an anteflexion effect on the torso are counteracted). At toe off, however, differences between the three columns were found: The lateral column (and to a less extent the intermediate column) showed unilateral activity to counteract the forthcoming lateroflexion of the torso.

The hypothesis that pelvic rotations determine ILBM activity was falsified. At the moment the pelvis rotates with the right side upwards, activity of the <u>right</u> ILBM is expected to keep the spine upright. However <u>left</u> ILBM activity was found, and thus ILBM activity is rather related to laterorotations of the thorax, which mirror the pelvic rotations closely (Thurston & Harris, 1983).

A phase lag is shown between the recorded EMG and the maximum force needed in the frontal plane as predicted by Cappozzo (1983). Probably also passive forces produced by lengthening of tissue come into play.

Therefore, in the **sixth** chapter a model is constructed, to estimate the changes in length of a line of action in the three columns at the different phases of a walking cycle.

The muscle length was found to be maximal at homolateral toe off, the phase in which the deliverance of a maximum force was necessary. Thus, it is likely that the maximum force needed to resist lateroflexion of the torso is produced by the passive lengthening of the ILBM and surrounding tissue.

In the fifth chapter only an indirect relationship was shown between pelvic rotation in the frontal plane and ILBM activity. In the **seventh** chapter it was studied whether ILBM activity is determined by pelvic tilt under isometric conditions. Pelvic tilt was created by raising one foot during standing.

The pelvic tilt increased linearly with increasing artificial leg length discrepancies. However, the RA-EMG appeared to increase only appreciably above a leg inequality of 35 mm (7.4 degrees of pelvic tilt). Such an increase was only found in the activity of the medial column and caudally in the intermediate column above the longer leg. Thus, apart from the functionally different columns a differentiation within the columns seems to be possible.

Apparently, the unilaterally recorded ILBM activity was required to counteract the contralateral force produced by the lengthening of structures above the short leg.

The influence of an artificial leg length discrepancy on pelvic rotations and ILBM activity was tested in the **eighth** chapter.

Based on the first seven chapters of this thesis it is predicted that during walking with unequal leg lengths, bilateral activity of the ILBM increases during heel strike of the raised limb. ILBM activity is required to counteract the increased forward flexion, because of a relatively increased deceleration of the raised limb.

Another prediction is that ILBM activity above the longer leg should increase to resist the contralateral passive force produced by the lengthening of structures above the short leg at toe off of this leg.

A problem in verifying the predictions is that the comparability between walking with and without a raised leg proved to be complicated by several factors. Walking with a raised leg produced a bilateral increase of the swing phase time and a decrease of the stance phase time, the stride times being still equal. Another problem was the minor influence of a raised leg on pelvic rotation in the frontal and sagittal plane during walking. A leg length inequality is probably largely compensated by changes in phase lengths of a stride as well as in changed knee flexion and pelvic translation along the Z-axis.

The RA-EMG of the three columns did not differ significantly between normal walking and walking with a raised leg, partly due to small changes and large standard deviations.

The activity time was influenced (activity was defined above 20 % of the maximum RA-EMG during walking). The first prediction was affirmed: Activity time increased bilaterally at heel strike of the raised leg. The second prediction could not be affirmed: activity time of the medial and intermediate columns was not influenced and instead of an increase of activity above the longer leg, the lateral column decreased its activity above the short leg at its toe off, probably because additional passive forces are produced by lengthening tissues above the short leg.

In spite of the small influence of a artificial leg length discrepancy on pelvic rotation, the activity time of the ILBM was changed, indicating that ILBM activity is less dependent of pelvic rotations and more related to thoracic movements.

Conclusion

The recruitment of the ILBM is strongly dependent on the plane in which the moments are required.

Sagittal plane: All three columns of the ILBM are recruited during extension. In submaximal forces the medial column shows relatively more activity in relationship to the maximal recorded activity than both other columns.

Frontal plane: The medial and caudal parts of the ILBM are more recruited during lateroflexion of the lower lumbar vertebrae and the lateral parts are more involved with lateroflexion of the upper lumbar vertebrae and the torso.

The method used to arrive at this conclusion is applicable in the prevention of back pain. The recruitment of back pain free subjects performing hazardous activities over years, recorded with the method used in this thesis, may indicate an optimal use of the ILBM. This optimal use directs training and environmental adaptations.

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10. Samenvatting

Er wordt veel onderzoek gedaan op het gebied van rugklachten. Desondanks is de oorzaak van rugklachten nog steeds onbekend. Het is ook onbekend hoe het ontstaan van rugklachten voorkomen kan worden. Om rugklachten te voorkomen worden nu enerzijds maatregelen genomen die de externe belasting verminderen (richtlijnen betreffende maximale tillasten en herinrichting van werkplekken), anderzijds worden pogingen ondernomen om de "mens aan te passen" door training en voorlichting. Dit wijst erop dat het probleem nogal complex is. Er is inderdaad een marginaal verschil in de prevalentie van rugklachten tussen lichte en zware beroepen (Dales et al., 1986). Dit is een aanwijzing dat het bijzonder zinvol kan zijn om de verschillen tussen mensen met en zonder rugklachten te bestuderen. Met name verschillen in het bewegingspatroon van mensen met en zonder rugklachten die al jaren risicohandelingen verrichten zijn daarbij interessant. Soms zal het vaststellen van de verschillen in grove motoriek al voldoende zijn om aanwijzingen te geven voor verbeteringen, maar soms zullen subtiele verschillen in het gebruik van de rugspieren bestudeerd moeten worden om tot die aanwijzingen te komen. Het electromyografisch vastgelegde patroon is bij het vaststellen van subtiele verschillen een belangrijk (zo niet het enige momenteel beschikbare) hulpmiddel.

In dit proefschrift is een methode beschreven waarmee het patroon van spieractiviteit laag in de rug (het deel waar zich de meeste klachten voordoen) gemeten kan worden tijdens het uitvoeren van risicohandelingen. Door deze methode in de toekomst toe te passen op mensen, die geen rugklachten hebben en de risicohandeling al jaren verrichten kunnen gegevens worden verkregen over een optimaal spiergebruik. Dit optimale spiergebruik kan geleerd worden aan nieuwkomers in het beroep. Daarnaast zal ook in een aantal gevallen de werkomgeving aangepast moeten worden om het optimale spiergebruik mogelijk te maken (dat kan ook een vermindering in de externe belasting inhouden).

De oorspronkelijke vraagstelling voorafgaand aan dit onderzoek was: 'Hoe vindt de musculaire stabilisatie van de wervelkolom plaats in verschillende belastende omstandigheden?'. Een gedeelte van deze belastende omstandigheden kunnen op de lange duur rugklachten veroorzaken, zoals het leveren van maximale krachten en het staan en lopen met een beenlengteverschil.

In **hoofdstuk twee** worden enkele recente bijdragen van de literatuur betreffende de anatomie van de lumbale rugspier beschreven. In het bijzonder wordt daarbij aandacht geschonken aan het werk van Bogduk en Bustami. In het lumbale deel kunnen morfologisch grofweg drie entiteiten onderscheiden worden in de intrinsieke rugspieren (zie fig 10.1): mediaal de m.multifidus, intermediair de m.longissimus en lateraal de m.iliocostalis lumborum. Het was onbekend of deze drie delen ook functioneel te onderscheiden waren. In een pilotstudie (Dofferhoff & Vink, 1985) werd reeds een verschil in functie aangetoond tussen twee ervan: de m.multifidus en de m.iliocostalis lumborum.

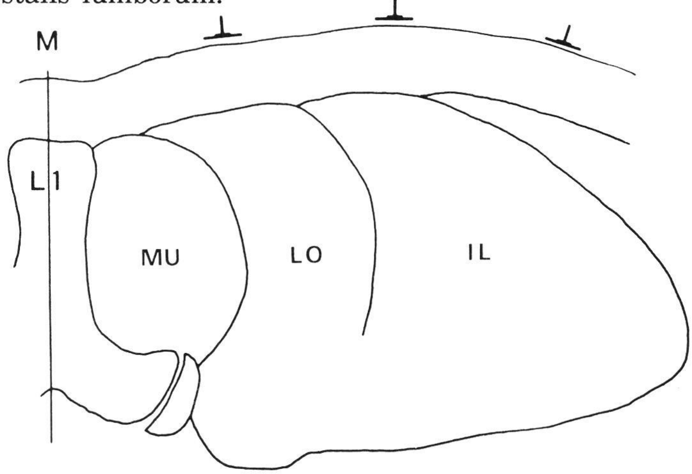

fig. 10.1. Drie morfologisch te onderscheiden entiteiten op eendwarsdoorsnede door de romp op het niveau van de eerste lumbale wervel zoals Jonsson (1970) dat aangaf. MU = m.multifidus, LO =m.longissimus, IL = m.iliocostalis lumborum.

In **hoofdstuk drie** van dit proefschrift is aangetoond dat verschillen in activiteit van drie meettechnisch te onderscheiden entiteiten, kan worden geregistreerd met 12 bipolaire oppervlakte electroden (zie fig 10.2). De drie entiteiten komen grofweg overeen met de hierboven gegeven anatomische indeling in drie onderdelen. Alleen bij de mediale kolom op L1 niveau zal zowel activiteit van de m.spinalis, m.multifidus als van de m.longissimus geregistreerd worden.

In dit hoofdstuk werd ook aangetoond dat met de beschikbare apparatuur een registratie met meer electroden dichter bij elkaar niet meer informatie oplevert, omdat de overspraak zal toenemen. Dat wil zeggen dat in het afgeleide signaal dan in te grote mate activiteit van omliggende spieren (of delen van spieren) mee wordt geregistreerd.

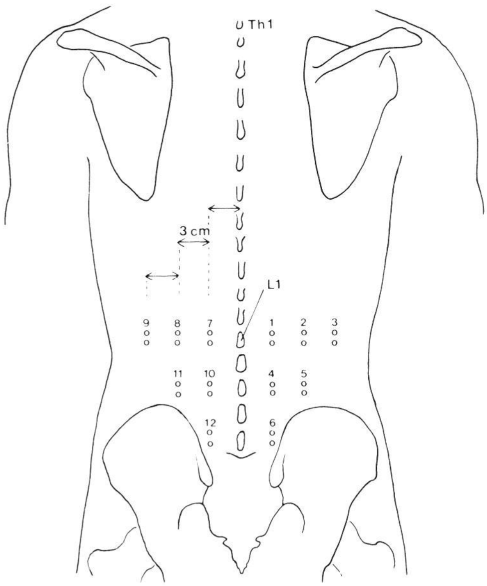

Fig. 10.2. De positie van de 12 bipolaire electrodes. De mediale kolom wordt geregistreerd met electrode 1, 4, 6, 7, 10 en 12, de intermediaire met 2, 5, 8 en 11 en de laterale met 3 en 9.

Naast gegevens over de lage rugspieren, leverde dit experiment algemene gegevens op inzake overspraak. Zo is aangetoond dat de crosscorrelatie-coefficient-functie tussen twee bipolaire oppervlakte electroden een bruikbare parameter is om de mate van overspraak te kwantificeren. Het absolute maximum in deze crosscorrelatie-coefficient-functie werd beïnvloed door diverse factoren, zoals statisch versus dynamische activiteit, de mate van activiteit, het ECG, de door apparatuur veroorzaakte ruis en brom, en de weerstand in het biologische materiaal. Het absolute maximum in de crosscorrelatie-coefficient-functie werd het meest beïnvloed door de afstand tussen de electrodes. Hoe groter de afstand tussen de electrodes des te lager werden de absolute maxima: de waarden waren 0.51, 0.22 en 0.15 voor electrodes die respectievelijk 30, 60 en 90 mm van elkaar af lagen. Het feit dat de andere factoren ook de overspraak beïnvloeden geeft aan dat deze factoren beheerst moeten worden, wanneer twee EMG signalen vergeleken worden. Dit houdt bijvoorbeeld in dat de invloed van deze factoren tot een minimum beperkt moet worden, of situaties vergeleken moeten worden, waarin de invloed evenveel of bekend is.

In het **vierde hoofdstuk** is EMG geregistreerd onder isometrische condities bij krachten variërend van 10 tot 100 % van de vrijwillige maximale krachtleverantie. De proefpersonen moesten een extensie van het bovenlichaam tegen weerstand uitvoeren in rechtop staande houding.

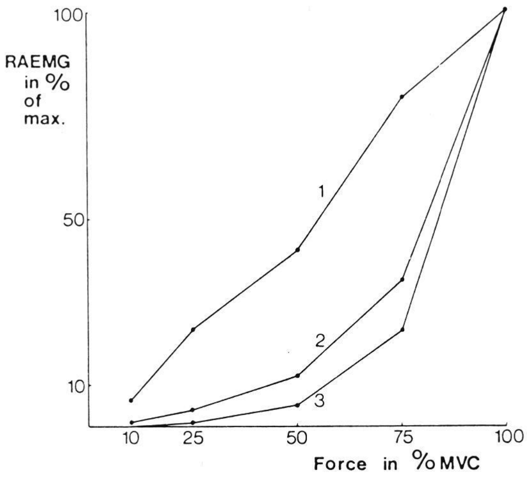

Fig. 10.3 De toename in RA-EMG voor drie kolommen op L1 niveau als functie van hun toename in kracht, beide uitgedrukt in percentages van het maximum. De nummers bij de lijnen corresponderen met de electrodenummers in fig. 10.2.

Tussen de drie entiteiten werden de volgende functionele verschillen gevonden (zie fig. 10.3): het RA-EMG van de mediale kolom nam vrijwel lineair toe met de extern gemeten kracht, terwijl het RA-EMG van de intermediaire kolom pas bij hogere krachtniveau's duidelijk toenam (RA-EMG is de som van in dit geval een seconde gelijkgericht (absoluut) EMG). Bij de laterale kolom was dit fenomeen nog meer uitgesproken. Binnen een kolom tussen de diverse lumbale niveau's werden geen verschillen in functie gevonden. Ook vermoeidheid en de volgorde van krachtsleverantie hadden weinig invloed op de vorm van de EMG-kracht relatie.

Wanneer het EMG niet gerelateerd wordt aan de maximale kracht, worden geen verschillen gevonden en lijkt het of de drie kolommen uniform gerecruteerd worden, een grote variatie in krachtniveau's is dus nodig om de verschillen tot uiting te laten komen. Tijdens staan en zitten vonden Andersson et al. (1974) geen significante verschillen tussen de drie kolommen omdat verschillende krachtniveau's niet gerelateerd werden aan elkaar. Bij de in dit proefschrift beschreven resultaten verschillen de drie kolommen wel significant. Bij submaximale extensie tegen weerstand vertoont de mediale kolom meer activiteit ten opzichte van zijn maximum dan de beide andere kolommen. De verklaring voor dit fenomeen ontbreekt nog, wellicht speelt de lengte van de spiervezels en het type spiervezel hierin een rol.

In **hoofdstuk vijf** zijn functionele verschillen tussen de drie kolommen tijdens lopen beschreven. Tijdens het plaatsen van de hiel op de grond werden nog geen verschillen gevonden: alle drie de kolommen waren bilateraal actief om het dreigende vooroverbuigen van de romp onder invloed van het traagheidsmoment tegen te gaan. Tijdens het verbreken van het teencontact met de grond waren de verschillen wel aanwezig. De laterale kolom is dan unilateraal actief (en in mindere mate de intermediaire kolom) om lateroflexie van de romp te kompenseren.

De veronderstelling dat de rugspieractiviteit bepaald zou worden door de mate van bekkenrotaties in verticale vlakken is onjuist gebleken. Op het moment dat de rechter helft van het bekken omhoog beweegt, zou ook aan de rechterzijde activiteit van de rugspieren verwacht worden om de wervelkolom rechtop te houden. Maar de rugspieractiviteit werd juist links gemeten. Dit komt waarschijnlijk omdat laterorotatie van de romp moet worden tegengegaan. Deze laterorotatie loopt precies tegenovergesteld aan de bekkenrotatie (Thurston & Harris, 1983). Dit verklaarde waarom er toch een correlatie tussen bekkenrotatie in het frontale vlak en het RA-EMG gevonden werd.

Er is bij een kompensatie van de lateroflexie een faseverschil tussen het moment waarop EMG activiteit maximaal is en het moment waarop de meeste kracht nodig is volgens voorspellingen van Cappozzo (1983). Deze voorspelde en in het EMG ontbrekende kracht zou passief door rek van weefsel geleverd kunnen worden.

In **hoofdstuk zes** is een model gepresenteerd, waarin de lengte van de drie kolommen wordt berekend in de verschillende stadia van de loopcyclus.

Het blijkt dat de lengte het grootst is op het moment dat de kracht ook het grootst zou moeten zijn. De conclusie is derhalve dat de kracht, die nodig is voor het lateroflecterende moment, gedeeltelijk passief geleverd wordt door verlenging van spieren en omliggend weefsel.

In het **zevende hoofdstuk** is nagegaan of onder isometrische condities de rugspieractiviteit direct beïnvloed wordt door de bekkenstand in het frontale vlak. Uit hoofdstuk vier bleek dat tijdens lopen rugspieractiviteit slechts indirect gerelateerd is aan de bekkenrotatie in het frontale vlak.

De bekkenscheefstand is gecreëerd door plankjes onder een voet te plaatsen.

De bekkenscheefstand nam lineair toe met een toename van het kunstmatig beenlengteverschil. Het RA-EMG nam echter pas duidelijk toe boven een gemiddeld beenlengteverschil van 35 mm (7.4 graden bekkenscheefstand). Dit punt van toename vertoonde een sterke interindividuele variatie (van 15 mm tot 45 mm). De toename werd alleen geconstateerd in de mediale kolom en caudaal in de intermediaire kolom homolateraal boven het verlengde been. De caudale activiteit wijst erop dat er naast een functioneel onderscheid in drie kolommen ook een differentiatie binnen de kolommen mogelijk is.

De gemeten rugspieractiviteit kan verklaard worden doordat zij nodig is om de passief geleverde kracht door de verlengde structuren boven het korte been te kompenseren.

In **hoofdstuk acht** is de invloed van een kunstmatig beenlengteverschil op rugspieractiviteit en bekkenrotatie tijdens lopen onderzocht.

Op grond van de eerste vijf experimenten is te voorspellen dat bij het lopen met een kunstmatig beenlengteverschil de activiteit bilateraal in alle kolommen toeneemt tijdens het neerzetten van de verhoogde hiel, omdat door grotere deceleratie de romp meer naar voren zou buigen. De toename is bedoeld ten opzichte van lopen met gelijke beenlengten.

Voorts is te voorspellen dat bij een grotere bekkenscheefstand de activiteit van de rugspieren boven het verlengde been toeneemt tijdens lopen. Door verlenging van structuren boven het korte been zal de passief geleverde kracht namelijk toenemen, welke contralateraal gekompenseerd zal moeten worden.

Een komplikatie bij de toetsing was dat de vergelijkbaarheid tussen lopen met en zonder beenlengteverschil verstoord werd. De staptijdvariabelen werden namelijk ook beïnvloed door het beenlengteverschil. Bij een verhoging werden bilateraal de zwaaifasen langer en de standfasen korter, terwijl de totale schredetijd gelijk bleef. Een andere komplikatie was de kleine invloed van een beenlengteverschil op de bekkenscheefstand tijdens lopen. Blijkbaar wordt een beenlengteverschil grotendeels door veranderingen in bijvoorbeeld staptijdvariabelen, knieflexie en bekkentranslatie langs de Z-as opgevangen.

Het RA-EMG van de drie kolommen tijdens lopen met een kunstmatig beenlengteverschil verschilde niet significant met gewoon lopen, mede door de kleine verschillen en de grote spreiding. Door als drempelwaarde een arbitraire grens te trekken op 20 % van het maximale RA-EMG tijdens lopen, kan berekend worden wanneer een kolom al dan niet actief is. De duur van de activiteit werd wel beïnvloed door een beenlengteverschil. De - eerste voorspelling wordt bevestigd: De duur van de activiteit nam toe rondom hielkontakt van het verlengde been. De tweede voorspelling lijkt dan niet bevestigd te worden: De mediale en intermediaire kolommen hebben een ongeveer gelijke duur van activiteit met en zonder beenlengteverschil.

Echter, in plaats van toename van activiteit boven het verlengde been, neemt de duur van de activiteit van de laterale kolom af boven het verkorte been, wellicht omdat een deel van de kracht passief door verlenging geleverd wordt.

Dus ondanks de constatering dat de bekkenrotatie weinig werd beïnvloed door een kunstmatig beenlengteverschil, is er toch invloed op de duur van de rugspieractiviteit. Een gegeven dat net als in het derde experiment aangeeft dat de rugspieractiviteit minder afhankelijk is van de bekkenrotatie, en meer van de thoracale bewegingen tijdens lopen.

Conclusie

De activiteit van de rugspier is sterk afhankelijk van het vlak waarin de geleverde momenten liggen.

Sagittale vlak: Alle drie de delen van de intrinsieke lumbale rugspier zijn actief bij extensie. Bij sub-maximale extensie is de mediale kolom relatief meer actief ten opzichte van zijn maximale krachtsleverantie dan de laterale.

Frontale vlak: De mediale en caudale delen zijn meer actief bij lateroflexie van de onderste lumbale wervels en de laterale delen meer bij lateroflexie van de bovenste lumbale wervels en de romp.

De methode die tot bovenstaande conclusie leidde is geschikt om te gebruiken bij de preventie van rugklachten. Daarbij zullen factoren, die de cross-correlatie beïnvloeden beheerst moeten worden en naast het EMG signaal ook andere gegevens verzameld moeten worden, zoals kinematische gegevens of krachten. Deze andere gegevens zijn nodig om na te gaan hoe groot de passief geleverde kracht is. Een optimaal spiergebruik kan opgespoord worden door registratie van de patronen van rugspieractiviteit van rugklachtenvrije personen, die al jaren risicohandelingen verrichten. Wanneer dit optimale spiergebruik is vastgesteld, geeft dit in belangrijke mate richting aan de arbeidsplaatsverbetering en training.

(referenties zie pagina 97)

List with abbreviations

ALLD = artificial leg length discrepancy
CCCF = crosscorrelation-coefficient-function
cmmr = common mode rejection
E = RA-EMG in % of maximal EMG
ECG = electrocardiogram
EMG = electromyogram
F = force output in % of maximal force
FER = force to EMG relationship
ILBM = intrinsic lumbar back muscles
LHS = left heel strike
LLI = leg length inequality
LTO = left toe off
L1 = first lumbar vertebra
L3 = third lumbar vertebra
L5 = fifth lumbar vertebra
MVC = maximal voluntary contraction
QRS = part of the ECG with a large amplitude, caused by depolarisation of the ventrikels of the heart
RA-EMG = rectified and averaged EMG
RHS = right heel strike
RTO = right toe off
SD-EMG = standard deviation of the EMG
SEM = standard error of the mean
Th4 = fourth thoracic vertebra

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Curriculum vitae

De auteur van dit proefschrift werd in Den Haag geboren op 23 april 1957. Na het behalen van het Atheneum B diploma op 18 juni 1976, is hij in september 1976 gestart met de studie aan de Interfaculteit Lichamelijke Opvoeding (= de huidige faculteit der bewegingswetenschappen). Op 20 september 1979 behaalde hij het kandidaats diploma aan de Vrije Universiteit te Amsterdam.
In de doctoraalfase is het hoofdvak functionele anatomie en zijn de bijvakken orthopaedie en onderwijskunde gekozen. De belangrijkste onderwerpen in de doctoraalfase waren EMG-spectrumverschuivingen, een curriculum optimalisatie van het vak kinesiologie en een onderwijsleermethode functionele anatomie.
Tijdens de studie heeft hij 3 jaar een student-assistentschap vervuld, wat onder andere de organisatie van het vak antropobiologie inhield.
Op 28 maart 1983 werd het doctoraalexamen behaald. Voor het behalen van de bul werd de postdoctorale cursus gevolgd, die hem een eerste graads onderwijsbevoegdheid verschafte.
Vanaf 1 september 1982 tot 1 oktober 1987 was hij voor 17 uur per week benoemd als docent biologie van de mens aan de Chr. Academie voor Jeugdwelzijnswerk. Gedurende deze periode heeft hij een leerboek geschreven (Gezond blijven, P.Vink, De Tijdstroom, Lochem, 1986).
Vanaf 1 juli 1983 tot 1 juli 1988 is hij voor een halve weektaak aangesteld bij de vakgroep Anatomie van de Rijksuniversiteit te Leiden. Een groot deel van het werk in deze periode is beschreven in dit proefschrift.
Vanaf 1 september 1987 is hij medewerker onderzoek en beleidsontwikkeling bij de Stichting Arbouw. De aandachtsgebieden binnen deze functie zijn bewegingsapparaat, arbeidsfysiologie en verzuim.