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The use of non-amplitude components of the myoelectric signal in identifying differences in function between the low back injured and controls

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This article primarily reviews the assessment of the non-amplitude dependent components of the myoelectric signal in assessing lumbar function in the low back injured and how persons with low back injuries may differ from the non-injured.

Assessment of anomalies in function can loosely be categorized into 4 separate protocols: 1. Assessing the pattern of myoelectric activation, 2. Assessing the EMG frequency spectrum during fatigue, 3. Assessing EMG onset timing during movement and stability challenges and 4. Assessing myoelectric function with link-segment models and EMG assisted spinal models.

Assessing the EMG spectrum during fatigue can discriminate between populations; however, questions regarding across day repeatability limit its ability to identify change over time. The remaining techniques are relatively new, show statistically significant differences between the low back injured and normals and provide insight into aberrant spinal stability, motor control function and spinal loading. Their response to rehabilitation programs is largely unknown suggesting a need and avenue for future research.

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KEY WORDS: chiropractic, low back, injury, EMG.

Cet article vise à évaluer la fonction lombaire chez les personnes atteintes de blessures dans cette partie du dos par comparaison à la celle des personnes sans atteinte lombaire principalement à l'aide des éléments du signal électromyographique sans mouvement humain.

On pourrait évaluer les anomalies de fonctionnement à l'aide de quatre protocoles différents : 1. Évaluer le mode d'activation électromyographique. 2. Évaluer le spectre de fréquence de l'EMG au moment de la fatigue. 3. Évaluer le réglage initial de l'EMG lorsque la personne est immobile et lorsqu'elle est en mouvement. 4. Évaluer la fonction électromyographique en utilisant des modèles de segments liés et des modèles spinaux assistés par EMG.

L'évaluation du spectre de l'EMG en période de fatigue peut présenter une différence entre les groupes; cependant, les problèmes concernant la fidélité de l'appareil d'une journée à l'autre limitent sa capacité à distinguer les changements sur une longue période. Les autres techniques sont relativement nouvelles et présentent des différences statistiquement importantes entre les personnes ayant une blessure de la région lombaire et les personnes sans blessure. Elles nous permettent de mieux comprendre la stabilité aberrante de la colonne vertébrale, le contrôle de la motricité et la capacité de la colonne vertébrale de supporter des charges. On ignore la réponse des patients aux programmes de réadaptation, d'où le besoin d'éventuels projets de recherche.

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MOTS CLÉS : chiropratique, bas du dos, blessure, EMG.

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Introduction

The myoelectric (ME) activity of trunk musculature is commonly used in an attempt to assess dysfunction of the lumbar spine to determine if differences exist between low back pain sufferers and normals. The majority of assessments in the chiropractic field have focused on quantifying the ME amplitude differences between low back pain patients and control subjects. The rationale behind these investigations is to identify "spasm" or increased muscle activity in low back pain populations as a result of muscle splinting or aberrant neural control. Van Dieen et al[1] systematically reviewed the literature to determine whether myoelectric changes in low back pain sufferers can be explained by either the pain spasm pain model or the pain adaptation model. The pain spasm model postulates that pain causes increases in muscle activity in turn causing more pain. The pain adaptation model postulates that pain decreases myoelectric activity in muscles acting as agonists and increases activity in muscles acting as antagonists. They conclude that the research fits neither model, changes are task dependent, related to individual problems, and highly variable between individuals and possibly within individuals.

One probable reason studies to date have not consistently demonstrated differences in symmetry and amplitude between low back pain patients and healthy populations may be due to the many factors that modulate measured EMG activity level which are not related to the level of neural drive. Electrode placement, skin temperature, moisture, cutaneous fat distribution, muscle fibre type and size can all influence measured EMG activity level.[2] Non-homogeneity in these factors between sides of the body may relegate asymmetry in measured EMG activity to be the norm even though it is possible that bilateral muscles are contracting at equal intensities. With so many factors modulating EMG activity a large variation in EMG amplitude is seen[3] across subjects. A patient may have an elevated EMG level relative to their normal activation level whereas their EMG activity level may still be within a range considered normal. It also possible that not all patients with back pain have a condition that presents with an elevated EMG trunk muscle activity. It is then probable that when group averages are compared no differences exist across groups even though patients within a low back pain group may have an elevated EMG compared to what is normal or healthy for them.

The uncertainties and variability of these results regarding the measure of trunk muscle activation level as an outcome measure and spinal assessment tool suggest its use in a clinical setting may presently be limited and other EMG spinal assessment procedures should be investigated. The following four sections all use surface EMG measurements but assess different components of the EMG signal or use different processing techniques in an attempt to categorize and describe spinal function. The majority of these outcome measures and spinal assessment techniques are relatively new and the research is still ongoing. This review is not arguing that these assessments of spinal function have stronger research support than previous measures of EMG amplitude. In some instances the research is very sparse and much is still needed

It is the aim of this review to introduce to the chiropractor the various ways in which low back pain sufferers differ from controls during assessments of primarily non-amplitude components of the trunk musculature's myoelectric signal. Four sections will review primarily non-amplitude assessments of the myoelectric signal in the low back injured. Those being: 1.Assessing the pattern of myoelectric activation, 2. Assessing the EMG frequency spectrum during fatigue, 3. Assessing EMG onset timing during movement and stability challenges and 4. Assessing myoelectric function with link-segment models and EMG assisted spinal models

Literature search strategy

Pubmed was used to find articles using the key words: EMG, low back pain and EMG AND low back pain up to January 2003. A qualitative assessment of the abstract determined if the article was relevant to the four non-amplitude based measures of myoelectric signal in the low back injured. No standardized rating of the article occurred. Additional articles were found by bibliographic searches. As well, some articles were excluded if the material they presented was redundant or had been improved upon by the same researchers in future studies. Articles were also excluded if they dealt solely with amplitude based measures of the myoelectric signal.

EMG Assessment #1: Assessing the pattern of EMG activation

This method of assessing trunk muscle activity evaluates the shape (linear envelope) of the electromyogram's activation profile. This technique looks at the changes in the muscle activation level over time making it possible to compare the shape of the EMG linear envelope (activation profile) across subjects or within a subject to compare bilateral muscle group symmetry. Grabiner[4] found a greater degree of erector spinae bilateral asymmetry (both shape and amplitude) in a low back pain population ($n = 6$) compared with a control group during an isometric exertion. A similar difference between populations was found by Lehman[3] during dynamic flexion tasks. This study quantified the symmetry in the bilateral erector spinae (upper T9 and lower L3) EMG linear envelope using a cross correlation function which assesses the similarity between the left and right EMG waveforms. They found that the left and right lower erector spinae linear envelopes (activation profile) were less similar (correlated) in low back sufferers compared with normals. No differences in the symmetry of the upper erector spinae was seen between the two groups. This study is limited in its small sample size and its clinical utility. How these findings relate to other measures of dysfunction and whether these findings change with a rehabilitation program is unknown. Lu et al.[5] found greater differences in bilateral asymmetries, in the linear envelope of the erector spinae in a low back pain population during trunk movements and lifting tasks. However, after an 8 week treatment period, despite clinical improvement, no changes in asymmetry levels were shown. This lack of symmetry has also been shown when comparing agonists and antagonists on the ipsilateral trunk musculature. Hubley-Kozey and Vezina[6] compared the ensemble average profiles of the EMG linear envelopes from trunk flexors and extensors using pattern recognition methodology during supine leg raising tasks designed to challenge spinal stability. The authors concluded that a healthy control group coactivated all muscle groups with the same pattern of activation while the low back pain group showed a lack of synergistic co-activation.

One limitation to the assessment of bilateral symmetry of muscle groups is the possibility that the bilateral **myoelectric signal** may not truly represent the bilateral muscle activation level. The myoelectric activity recorded under EMG electrodes can be greatly influenced by small differences in electrode position. Therefore other portions of the same muscle group may be active to a different extent than the portion that is being measured. Therefore the measured symmetry may be different from the true activation symmetry of the overall muscle group.

In addition to looking at bilateral symmetry in muscle activation profiles researchers have investigated the activation levels of muscles in functional pairings. An example of this is seen in a study investigating the EMG activity coupling of the gluteus maximus muscle and contralateral latissimus dorsi muscle in patients with SacroIliac dysfunction.[7] Vleeming et al.[8] have demonstrated the anatomical fascial connection between these two muscles hypothesizing that this link provides stability for the SI joint. Mooney et al.[7] investigated the EMG activity of the these two functionally linked muscles in patients with SI dysfunction during rotational exertions and concluded that different muscle activity ratios were evident between the gluteus maximus and latissimus dorsi in the SI patients when compared with controls. The muscle activity relationship was also modified towards a normal standard over the course of a 2 month rehabilitation program. Unfortunately, no quantifiable ratios between the two muscles were reported, the sensitivity and specificity of this test was not done, nor was a statistical analysis performed comparing the two groups objectively. Nonetheless, this type of myoelectric assessment may hold promise in objectively documenting SI dysfunction. Research is needed to document a normal activation ratio between the gluteus maximus and contralateral latissimus dorsi. If an optimal muscle activation ratio is established for a variety of tasks clinician/researchers may be able to determine which exercises and therapies can best influence aberrant muscle activation ratios.

EMG Assessment #2: Assessing the EMG frequency spectrum during fatigue

EMG spectral parameter assessment refers to different ways of measuring and representing the frequency content of the raw EMG signal, which is composed of different frequencies between 10 and 500 Hz. One measure of the frequency content in a signal is the median frequency (i.e., the frequency of the EMG signal that divides the signal into two halves of equal power[9]). During isometric fatiguing contractions a compression of the power density spectrum of the EMG signal toward lower frequencies occurs. The rate of the decrease in the Median frequency (MF) provides an index of fatigue for the task performed.[10] This MF is measured using a mathematical technique called the Fast Fourier Transform (FFT). Wavelet analysis is another mathematical analysis which can be used to provide frequency information from a signal.

Unfortunately, the protocol for some of these studies to measure lumbar musculature fatigue is equipment intensive. In many investigations the protocol requires the pelvis and lower limbs to be stabilized and supported while the spine is held in a consistent position of neutral or thirty degrees flexion. The subject then exerts an extension force either against a pad behind them or against a chain that is secured to a vest they are wearing. Subjects perform a maximum voluntary contraction (MVC) to determine the amount of force they will exert during the fatiguing trials. The fatiguing trials require the subject to exert 40, 60 or 80% of their MVC for a period of 30 seconds. A rest period of 60 seconds occurs and the exertion is repeated for 10 seconds.

By comparing different measures of frequency and amplitude of the myoelectric signal (Initial median frequency, slope of the median frequency, root mean square of the amplitude of the signal) researchers have been successful in discriminating between low back pain symptomatic populations and pain free populations with sensitivity scores ranging from 76% to 88%.[9–17] Monitoring the spectral parameters of the EMG signal of the lumbar erector spinae, and multifidus during fatigue has shown superior discriminant validity than isometric strength measurements and range of motion assessments.[13]

While assessing EMG spectral parameters during fatigue has strong support for discriminant validity its ability to track changes during a rehabilitation program and its relationship to other outcome measures has been less well evaluated. Mannion et al.[17] found that over the course of three different therapies the Biering-Sorensen time to fatigue increased 18% but no change in the spectral parameters were seen. This is contrasted in the study by Roy et al.,[16] who, using an equipment-intensive protocol for assessing lumbar fatigue, found that over the course of a 4 week rehabilitation program participants showed an improvement in the spectral EMG measures (less of a decrease in the mean power frequency during the fatiguing contraction) of the lumbar musculature. This improvement in spectral parameters was also found at a one year follow up by Kankaanpaa et al.[18] in a chronic low back pain population following 12 weeks of active therapy.

Issues regarding the repeatability of spectral measurements of the myoelectric signal

Ideally, assessing the frequency changes in the myoelectric signal during a fatiguing task will identify patients with highly fatigable trunk extensors, irrespective of volition, and then provide a means of tracking improvements in muscle endurance over the course of a rehabilitation program. The success of such an assessment demands that the measures used to quantify fatigue be reproducible over time. A large variability in these measures would make it impossible to identify changes in muscle endurance and evaluate any changes to be due to the rehabilitation protocol. Unfortunately, assessing the frequency changes of the myoelectric signal to track improvement over time may be limited due to a large between day variability. The repeatability measure is often high (ICCs > .7) for the Initial Median Frequency[19,20,22–24] for various lumbar muscle sites. Combining muscle sites and controlling for posture improves reproducibility[20,24] the measure of the rate of fatigue (the slope or decrease in the median frequency over the contraction time) often shows variable repeatability results. Some studies show poor repeatability[19–21] while others using very similar test protocols find acceptable repeatability.[22–24] Use of the change in the root mean square amplitude consistently showed less repeatability than the frequency measures. It should also be noted that even when Intra-Class Correlation coefficients are high (> .8) one should also look at the Smallest Detectable Difference (SDD) which may suggest that the use of the assessment is limited. The SDD essentially says how much change in a measure must occur to say that that change is not due to chance. The studies which have calculated this suggest that changes in median frequency must be greater than 30%–100%[21,23] even when ICCs approach or exceed 0.9. It also important to control for posture as muscle length can influence spectral measures of myoelectric signal as well as amplitude measures.[20]

In addition to an increased fatigability of the erector spinae in the low back injured as measured by spectral parameters researchers have also shown an increased fatigability of the gluteus maximus relative to healthy controls.[25] This increased fatigability of the gluteus maximus may compromise SI joint stability as proposed by the work of Vleeming[8] mentioned in an earlier section of this review. If the coupling relationship between the gluteus maximus and latissimus dorsi is disrupted due to fatigue, SI stability may be jeopardized. The increased fatigue (or reduced back muscle endurance) of the erector spinae musculature has also been correlated with an increased inhibition of the knee extensors in golfers with chronic low back pain.[26]

In conclusion, measures of the frequency content of the myoelectric signal during trunk muscle isometric contractions appear sensitive and specific in identifying those with low back pain when a volitional effort is given. However, due to the large variability as measured with the Smallest Detectable Difference (SDD) its use in tracking improvements of an endurance program seem limited. It must be questioned whether myoelectric spectral assessments (which are equipment and time demanding) provide any more information in tracking improvement than a simple trunk holding test or other back endurance tests which measure time to fatigue or subject perception using Borg scales. It should be noted that even those repeatability studies which have found the highest ICCs with the most stringent experimental protocols conclude that the myoelectric frequency measures are not reliable enough to be interpreted at an individual level (24), maintaining that the interpretation of these EMG indices must be limited to group tendencies.

EMG Assessment #3:
Assessing EMG onset timing during movement and stability challenges

Two different biomechanical assessment techniques fall under this category. The first records the timing of trunk muscle recruitment during a voluntary movement of a limb. The second measures the latency of the response of selected trunk muscles during sudden unexpected loading of the spine. Both techniques are attempting to assess the motor control capabilities of the participant tested. The first technique is essentially looking for muscle inhibition or aberrant motor control strategies during a simple movement. The second technique attempts to de-stabilize the spine and record how the muscles respond in an attempt to achieve stability following the de-stabilization.

One variation of the first technique is an investigation into the postulated ideal muscle recruitment pattern during prone leg extension (PLE). It has been hypothesized that during normal PLE a typical and consistent pattern of muscle activation order should occur.[27] According to this theory, an aberrant temporal recruitment pattern decreases the stability of the pelvis during gait and thus hinders the body's mechanical efficiency. The typical variation of this recruitment pattern is described as the lower crossed syndrome. Theoretically, it includes a tightness of the erector spinae musculature and hamstrings and a weakness or inhibition of the abdominals and gluteus maximus muscles. Theoretically, lower crossed syndrome presents during a prone leg extension as a delay in the gluteus maximus recruitment. It has been hypothesized that the poorest recruitment pattern occurs when the gluteus muscle activation is delayed and the hip extension is achieved by forward pelvic tilt and hyperlordosis of the lumbar spine.

Bullock-Saxton[27] tested the temporal recruitment pattern during a PLE test, comparing patients with previous ankle sprains (within last 4 months) to a healthy control group. The injured group consisted of 20 men and the control group of 11 healthy men. Their study showed a greater delay in the activation of the gluteus maximus during prone leg extension in the ankle sprain group. The control group's activation pattern revealed that the activation of all muscles tested was almost simultaneous with gluteus maximus typically the last to become active.

The second study[28] to evaluate the muscle recruitment order during the PLE similarly found a consistent order of activation and found the gluteus maximus to be the last muscle activated. In contrast to the Bullock-Saxton study, the authors report an ordering of activation in a healthy group. Prior to the initiation of movement 6 muscles were activated in the following order: contralateral rectus abdominis, ipsilateral rectus abdominis, rectus femoris, ipsilateral lumbar erector spinae, contralateral lumbar erector spinae and semitendinosis. The tensor facia latae was recruited almost simulataneously with the onset of movement, followed by the gluteus maximus. However, when the means and standard deviations (expressed as a percentage of the movement cycle) are compared for the onsets of the ipsilateral erector spinae (mean = 13.91, SD = 10.97), contralateral erector spinae (mean = 17.27, SD = 12.86) and hamstrings (mean = 17.61, SD = 13.04) there is a great deal of overlap and they occur very close in time (amount expressed in seconds not known). This proximity in time may be identical to what the Bullock-Saxton study found but described as "almost simultaeneous". Again, with the large overlap in the Vogt and Banzer[28] study, it is possible that some of the muscles came on in a different order even though statistically a significant difference between the muscle onsets was found. The differences between the conclusions may have been due to collecting similar data, analyzing it slightly differently and subsequently finding a different conclusion.

Despite their differences, both studies provide support that a normal pattern of activation in healthy subjects sees the hamstrings and bilateral erector spinae fire almost simulataenously with the gluteus maximus the last to become active. Deviation from these consistent patterns of activation, especially an increase in the delayed firing of Gluteus Maximus suggests pathology. In addition, the anticipatory activity of the trunk muscles suggests the importance of the motor control system in providing adequate stability. This anticipation is seen in the following biomechanical assessment of lumbar function that specifically looks at the muscle recruitment during arm flexion.

Sufficient spinal stability and the motor control of spinal stability are considered necessary in the prevention of low back injury. Insufficient spinal stability during activities of daily living may predispose individuals to low back pain. In vivo measurement of spinal stability is necessary in evaluating the motor control properties of the stabilizing system of the spine. Hodges[29,30] documented the muscle recruitment patterns during flexion, abduction and extension of the arm during upright stance in low back pain and symptom free controls. In normals, the transverse abdominis, as measured with indwelling electrodes, shows anticipatory activity before the onset of movement. This is contrasted with a delayed activation time in subjects with low back pain suggesting motor control deficiencies in these patients and possibly decreased spinal stability. This delay in activation has also been recorded in low back injured patients in the internal oblique and the transverse abdominis during fast and intermediate movement of the upper limb but not during slow movements.[31] This delay in the transverse abdominis has also been shown to occur with movements of the lower limb in subjects with low back injuries.[32]

This technique is an excellent example of applying biomechanical experimental techniques to rehabilitation and injury prevention programs. Training programs designed to recruit the Transverse Abdominis and other trunk muscles have resulted in decreases in low back pain recurrence over 3 years from 75% in a control group to 35%.[33] These exercises may be necessary in retraining the motor control system to stabilize the spine. This stability is demonstrated during an abdominal bracing exercise which recruits the transverse abdominis and has been shown to decrease the laxity and increase the stability of the SI joint.[34]

The second method of evaluating the motor control of the lumbar spine is via measuring the latency response of the erector spinae to sudden unexpected loading.[35–39] This technique essentially disturbs the equilibrium of the spinal system and records the time it takes the trunk musculature to react to this instability and provide sufficient muscular stabilization. Patients with low back injury have been shown to have an increased latency response time to sudden loading using various techniques.[35–39] Two techniques are typically used. One sees the participant blindfolded holding a board in their hands. A weighted ball is dropped onto the board causing the participant to unexpectedly move forward. The time from the ball strike to the muscular activation is considered the latency response.[36,37,39] The second technique finds the participant seated and wearing a shoulder harness with cords attached from the harness to a secured object. The cords connected to the fixed object are composed of two parts that are joined by an electromagnet. This electromagnet can be shut off thus freeing the participant from their attachment to a fixed object.[35,38] With this technique no external force is added, rather the participant attempts to pull against the attached cord. While pulling against the secured cord the electromagnet can be shut off causing the participant to lurch in the direction that they were pulling. If pulling forward their spine would suddenly flex and the erector spinae would be required to become activated to re-stabilize the spine. This type of technique can be referred to as a quick release. Using this technique Radebold et al.[35,38] documented delays of the trunk musculature response to sudden loading in patients with low back injury compared with healthy controls (in all three planes of movement). One small study found that physical therapy treatment was effective in changing the erector spinae's latency response to sudden loading.[37] The sensitivity and specificity of its discriminant ability is unknown.

EMG Assessment #4: Assessing myoelectric function with link-segment models and EMG assisted spinal models

The majority of the assessment techniques previously reviewed have solely focused on the electromyographic measures of skeletal muscle during a variety of tasks. Little information during these spinal assessments is provided in regards to the kinematic (movement characteristics) or kinetic (joint loads or bending moments occurring about the low back) properties of low back injured populations. In order to calculate joint kinetic information from a subject, kinematic information (acquired from video, optoelectronic equipment or electromagnetic positional sensors) and force information (acquired from force plates or other strain gauges) must be combined with a link segment model to calculate joint reaction forces and moments. This is a time and equipment intensive task but this investment of resources does benefit the scientist by providing greater and more detailed information about how the lumbar spine is being loaded. Having load information helps explain the acquired EMG information. EMG data can be understood in the context of the loading requirements placed on the muscular system as catalogued by the link segment model. A link segment model allows the researcher to calculate the moment about the lumbar spine. Knowing how demanding a task is can help explain why differences might exist in the amount of myoelectric activity measured between two different subjects or between two different tasks. Additionally, the information from a link segment model can be inputed into an EMG assisted spine model to allow for the partitioning of the reaction moment between the force components provided by the disc, ligament strain, and active-muscle contraction. This allows the researcher to find each tissues contribution of shear and compression.

A dynamic spinal model was used by Marras et al.[40] to compare the trunk loading in patients with low back pain and controls during lifting tasks. This biological spine model uses information gathered from a link segment model and incorporates the EMG activity measured to create a dynamic spinal model. Not only is the moment about the lumbar spine calculated from the external demands of the task (as measured by the link segment model) but the loads applied to the spine via the musculature are also calculated which takes into account the influence of muscle co-contraction. The authors found that patients with low back pain had 26% more compression and 75% more lateral shear compared with controls. Patients modified their lifting techniques to minimize the torque created by external loads. However, despite the benefit this should cause for decreasing spinal loads, the patients offset this benefit by co-contracting the spinal musculature in order to ensure sufficient spinal stability. This co-contraction has also been seen in static tasks.[41]

Van Dieen et al.[42] found similar results when comparing muscle activation and Moment ratios of trunk antagonist activity over agonist activity and lumbar erector spinae (LES) over thoracic erector spinae (TES) during trunk movements and ramped exertions. The authors found that during sagittal plane ramp exertions a tendency toward higher antagonistic moments in patients was present. The moment contribution of the LES relative to that of the TES was significantly higher in the patient group in both motion and ramp contractions. The Moment results were similar to the electromyographic ratios. The electromyographic ratios of the antagonist over the agonists was signficantly higher in the patient group during movement trials (but not ramp contractions) as were the ratios of the LES activity relative to the TES activity.

In an apparent contradiction, Lariviere et al.[43] found an increased activation of the *left* thoracic erector spinae relative to the the lower erector spinae, a link segment model was used to calculate the external trunk moment. This increased activity was not seen on the right. The reason for the difference between the two studies is not explained and may be related to differences in the type of injury sustained by the two groups. Nor is it explained why a difference was only found on the left. It does suggest that differences in patient function may be patient specific. Not all patients with back pain may have the same functional differences.

By calculating the external loads on the spine the trunk myoelectric activity can be better explained. For example, since the moments about L5/S1 were the same for both groups in the Lariviere study, the difference in the EMG activity can not be accounted for by different demands on the musculoskeletal system. The subjects were performing the same tasks with identical requirements yet low back pain sufferers had aberrant muscle activation. The biomechanical model in the van Dieen[42] study helped show that the lower erector spinae contribute more to the internal moment production supporting the idea that the increase in muscle activity is functional in that it helps increase stability or compensate for a loss of stability elsewhere. Biomechanical models provide insight into the aberrant behavior of trunk muscle function. The additional moment information helps explain the EMG activity in the case of a link segment model, while an EMG assisted spine model permits the partitioning of forces amongst different muscles and passive structures required to balance the external moment calculated by a link segment model. However, this information comes at the expense of equipment and time resources.

Summary of EMG techniques of spinal assessment

Simple assessment of the amplitude of the EMG signal appears ineffective in discriminating low back pain patients from controls. How the myoelectric signal changes over the course of treatment and the relationship of these changes to other outcome measures is unknown. This review looked at four different means to evaluate lumbar function. While some of the procedures have not been excessively researched, the investigations of complex characteristics of lumbar function are beginning to demonstrate the ability to delineate and document differences between low back pain sufferers and healthy controls.

Low back pain sufferers appear to have a decrease in objectively measured spinal extensor endurance (as measured with spectral parameters), aberrant coupling during dynamic movements between bilateral muscle groups and functional pairings (e.g. the gluteus maximus: latissimus dorsi relationship), an altered response to sudden loading and altered stability responses. The greatest limitation to this research is that the relationship between these assessments and other outcome measures is unknown as is their response to treatment. Additionally, these studies inherently assume that each person with low back pain would demonstrate aberrant functioning as measured by the assessment technique. Whereas it may be more likely that low back injury causes different aberrations in function. These assessment techniques would then permit an ability to describe a patient's dysfunction in terms of quantifiable biomechanical data. For example, following a series of tests it may be found that a patient has adequate spinal endurance and normal symmetry between the gluteus maximus and the latissimus dorsi. However, the patient may also present with delayed muscle activation to sudden loading and muscle activation timing differences during abdominal exercises. By categorizing patients in terms of these functional deficits it may be possible to design treatment regimes to specifically correct these dysfunctions. These assessments may also be a means of delineating possible mechanisms of spinal manipulation or other treatment options.

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