0008-3194/89/82–92/$2.00/©JCCA 1989

Review Article

Recent advances in lumbar mechanics with relevance to clinicians

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Perhaps the singular, most important impediment to universal recognition of chiropractic as a legitimate and mature health care alternative is the absence of a developed mechanical knowledge of chiropractic techniques. The purpose of this review paper was to describe, and to some extent critique, some recent research pertaining to mechanics of the lumbar spine and to illustrate the relevance to clinical chiropractic. Specific contentious issues addressed include discussion of the mechanical relationship of intra-abdominal pressure, the lumbodorsal fascia, muscle-ligament interplay and the abdominal musculature with the lumbar spine. Directions or future research are proposed given the pressing need to provide a rationale for, and explanation of, specific chiropractic treatment. (JCCA 1989; 33(2): 82–92)

KEY WORDS: chiropractic, mechanics, lumbar spine, manipulation.

Peut-être que l'empêchement le plus important et le plus significatif pour la reconnaissance universelle de la chiropratique en tant que médecine alternative légitime et de grande maturité est l'absence de connaissances mécaniques pleinement élaborées des techniques chiropratiques. La présente communication critique avait pour objectif de décrire et, dans une certaine mesure, d'examiner du point de vue critique des recherches récentes sur la mécanique de l'épine dorsale lombaire et d'en illustrer par le fait même l'importance pour la chiropratique clinique. Parmi les questions litigieuses précises dont traite la présente communication figure une description des rapports mécaniques de la pression intra-abdominale, de l'aponévrose lombodorsale, des interactions entre les muscles et les ligaments et de la musculature abdominale par rapport à l'épine dorsale lombaire. Des lignes directrices pour la recherche future sont proposées, compte tenu du besoin pressant de formuler la raison d'être et une explication de la thérapeutique chiropratique proprement dite. (JCCA 1989; 33(2): 82–92)

MOTS CLEFS: chiropraxie, mécaniques, épine dorsale lombaire, manipulation.

Introduction

The legitimization of Chiropractic to significant portions of the public was dramatically increased with the introduction of various legislation including OHIP coverage in the early seventies. The single, remaining major impedance to universal recognition of chiropractic as a legitimate and mature health care alternative is the absence of a developed knowledge of the mechanisms of chiropractic techniques. Recent efforts of the chiropractic profession have been directed towards developing a substantive research program to provide rationale, and scientific support for chiropractic techniques. The objective of much ongoing research throughout the world is to increase understanding of the mechanical function of the lumbar spine. I have been asked in this review to describe some recent research of my own, and of others, pertaining to mechanics of the lumbar spine and to forge a link with clinical chiropractic.

How often have you been asked, "What role has the abdominal musculature to play in supporting the low back?" This remains a controversial issue in the literature. Another issue which remains open to debate questions the mechanical contributions provided by the lumbodorsal fascia. Even selection of the most efficient pattern of muscle-ligament interplay to support the lumbar spine during various loading activities is not universally agreed upon. Indeed, there are many other unanswered basic questions regarding lumbar function. Perhaps the most expedient approach to resolve such queries is to examine the mechanical behaviour of lumbar components by obtaining estimates of the magnitude and time history of loads in the involved tissues. Most of us would agree that various types of mechanical forces and torques precipitate lumbar dysfunction, and that Chiropractic utilizes the application of force to restore function. However, the fact remains that there is no easily implemented, non-invasive method to measure internal loads or

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stresses. In the absence of a direct measurement method, estimation of internal loads by mathematical modelling techniques remains the only tenable option. Most of the discussion in this review is based on results obtained from a mathematical model of the lumbar spine. Great efforts were directed towards incorporating anatomical detail into the computer model while data to drive the model were, in fact, biological signals collected from intact, individual, living humans to achieve an acceptable level of mechanical and biological fidelity. In a nutshell, the model does not perform simulations but rather utilizes muscle activation (EMG) signals from the abdominal and lumbar musculature of the subject, and measures of vertebral position to set the geometry of the low back tissues which enable estimates of their load-histories during the performance of a dynamic task. Readers who are interested in the fine details of the model, for specific tissues during various tasks, are advised to consult McGill and Norman[1,2,3].

At this stage, a simple review of lumbar muscle morphology will provide perspective necessary for discussion. Most models in the past have grossly simplified the extensor tissues (both muscle and ligament) to a single representative force generator connecting adjacent spinous processes (see figure 1). Further, estimates of muscle size (hence potential to produce force) were nearly always obtained from cadaveric material (e.g. Farfan[4]), which have been shown to be gross underestimates of muscle sizes obtained from transverse CT scans (reported by Reid and Costigan[5] and McGill[6] et al.). Clearly, in hindsight one could not expect such a gross simplification of the lumbar system to provide insight into mechanical function. A comprehensive view of lumbar mechanics is only possible from a reasonable representation of the anatomy. Excellent anatomical work by Langenberg[7] and Macintosh and Bogduk[8] provided clear descriptions of the primary extensors of longissimus thoracis and iliocostalis lumborum. One can observe from figure 2, that the fibre arrangement of these dominant extensors drastically change the mechanics of the joint by contributing a large shearing component together with increased moment arms to minimize joint compressive load through mechanical advantage. In addition, this laminated architecture facilitates a much larger muscle cross-sectional area to contribute to extensor moment production when compared to the simplified single equivalent representation that was assumed to be parallel to

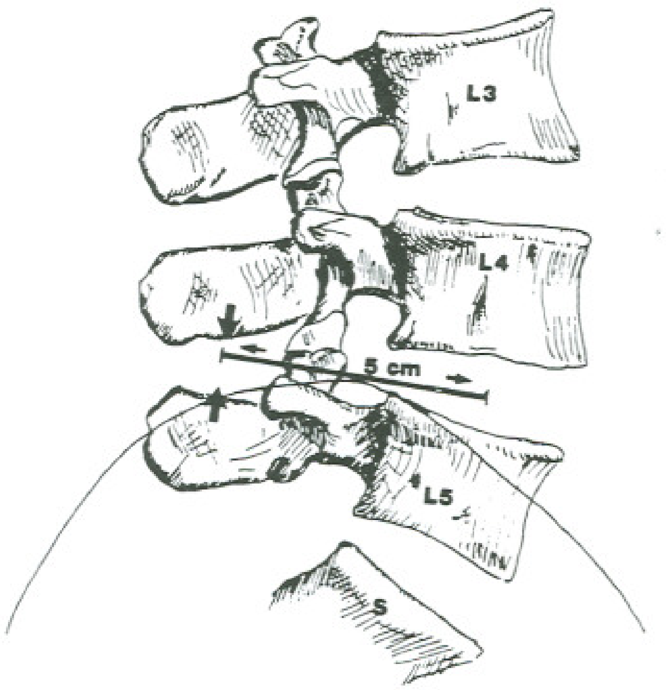


Figure 1 An example of the very simple, single equivalent force vector used to represent all of the extensor tissues of the lumbar spine. This model has been successfully used in industrial task analysis but should not be expected to provide insight into lumbar mechanics.

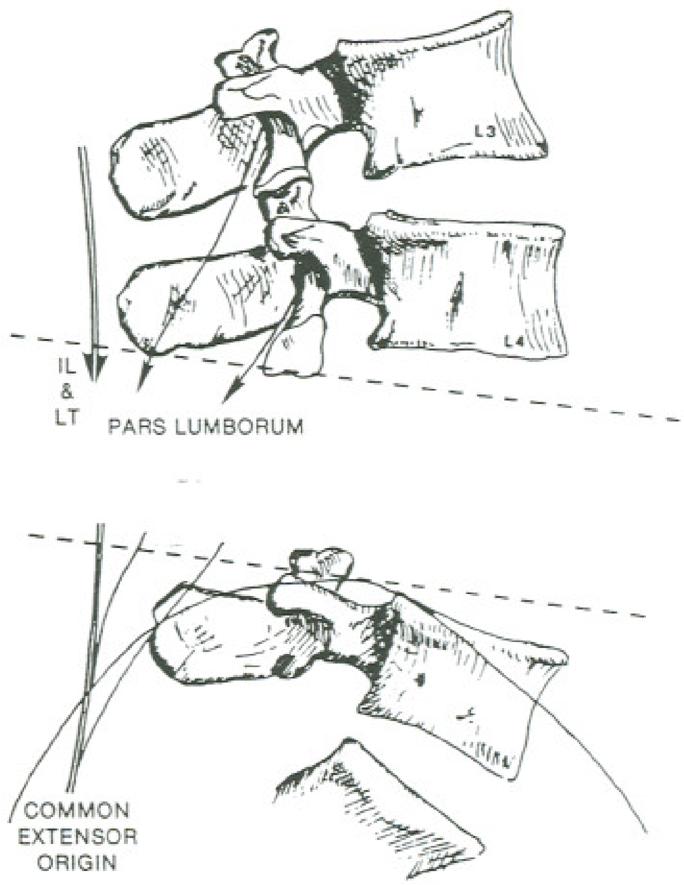


Figure 2 The architecture of the primary extensors reveals the shearing component of the pars lumborum fibres of longissimus thoracis and iliocostalis lumborum. While these laminae are quite close to the disc the tendons of the pars thoracis fibres contribute primarily to compression but generate extensor moment with a much greater mechanical advantage.

the compressive axis. In fact, during formulation of the model the architecture of 50 lumbar muscles including the abdominals and the 11 ligaments that cross the lumbar joints (excluding L5/S1 were examined and this detail was incorporated. Examination of model output was interesting from several perspectives. For example, ligament and muscular forces were shown to greatly affect facet load. During extreme flexion, strain of the interspinous ligament imposes an anterior shear on the superior vertebra (increasing facet load) while activity in pars lumborum components of iliocostalis lumborum and longissimus thoracis, associated with extension, create a posterior shear to unload the facet joint. Examination of axial twisting is another example where an accurate documentation of the mechanical history would be impossible if the the detailed geometry of the abdominal obliques, latissimus dorsi, and ligaments were ignored.

Several hypotheses regarding the mechanical roles of various tissues in the low back have been investigated recently, for example, the role of intraabdominal pressure and abdominal muscle activity in lumbar support. Knowledge of tissue load histories obtained from modelling techniques has been coupled with additional experimental evidence to enable assessment of the various mechanical proposals. Permit me to briefly relate the history of the marriage between analytical-modelling and experimental approaches to set the stage for the following presentation. Very simple anatomic models have been utilized for the last 3 or 4 decades to estimate tissue loads. Unfortunately, tissue load estimates from these models presented a paradox to the researcher. Joint compressive loads, that resulted from muscle, ligament and external forces, were predicted that exceeded vertebral end plate failure tolerance during quite reasonable lifts, yet laboratory subjects were able to complete the task uninjured. This seeming anomaly, was addressed from two perspectives. Perhaps the tissue tolerance data describing the strength of the vertebrae were too low, in other words, intact invivo vertebrae are better able to withstand compressive load than prepared cadaveric material under laboratory conditions. This may be possible considering the recent data of Miller et al[9]. Certainly, in highly trained individuals, evidence of increased vertebral mineral density has been presented by Granhed et al.[10], suggesting higher loading tolerances in very active people. However, given this explanation to account for the paradox resulting from the output of simple models, the popular direction of research effort was to formulate mechanically based hypotheses to explain how these large estimates of joint load could be reduced to tolerable levels. Hence the abdominals, for example, were implicated to work in various ways to reduce back loads. One hypothesis suggested that the abdominals assisted in the buildup of intra abdominal pressure which performed the dual function of exerting hydraulic forces upwards on the diaphragm to alleviate compressive loads on the lumbar column as well as contributing extensor moment to assist in unloading of extensor tissues. Since the extensor tissues are the major contributor to joint compression, any strategy to assist the extensors would minimize the compressive penalty imposed on the joint. Another thought provoking hypothesis speculated that the abdominals assisted in lumbar extension by transmitting lateral forces to the lumbodorsal fascia (LDF) (i.e. Gracovetsky et al[11]). It was argued that extensor contributions provided by the LDF would impose less compression on the lumbar joints by virtue of its larger distance posterior to the joint or improved mechanical advantage over the extensor musculature. Unfortunately, recent experimental evidence, coupled with analytical evidence from a detailed anatomic model have shown these claims to be extremely contentious; argument will be developed later in this review.

There is no need to explain the relationship between lumbar function and chiropractic to the readers of this journal. However, a thorough understanding of recent findings regarding lumbar function constitutes a step towards providing substantive, scientific support for the expanding role of Chiropractic as a health care profession. The following section is a synopsis of some of our findings, integrated with additional experimental evidence, which have implications on lumbar mechanics.

Estimates of specific tissue loads and implications on lumbar mechanics

There is no simple explanation to describe how the complex mechanical structure of the spine is able to successfully respond to the load demands that are part of daily living. Only knowledge of loads on the many individual component tissues will enable evaluation of the many controversies associated with spinal mechanics. Estimates of individual tissue loads were obtained from the anatomically complex model (McGill and Norman[1]) of which a schematic is shown in figure 3. The model was driven from biological signals collected from healthy subjects.

The measurement of lordosis is an appropriate beginning to the analysis. Separating trunk rotation about the hip from that about vertebral joints, and reporting trunk kinematics in this way, is critical for evaluation of lumbar biomechanics. Let us consider lordosis throughout a squat style lift performed by subjects in our laboratory. Moderate lordosis was preserved throughout the lift in all of our subjects suggesting that ligaments were not strained in flexion. In fact, this is demonstrated by most Olympic competitors. International coaches state that occasionally they encounter lifters who do not maintain lordosis during lifting, however these individuals are unable to lift competitive loads. A natural selection process takes place whereby these lifters are eliminated from high level competition by injury or simply cannot lift to a competitive level. This observation can be partially explained using knowledge of disc structure. Disc models suggest that the annulus is most resilient against compressive load when its "neutral" posture, neither flexed nor extended, is preserved. For example, the model of Hickey and Hukins[12] demonstrated that the neutral position is the only position where all annular fibres bear equivalent load and stress is equalized. Once compression is applied to a flexed disc, anterior annulus fibres become progressively disabled, and transfer their share of supporting responsibility to the posterior annulus. This is perhaps one suggestion to explain the preservation of lordosis during heavy lifts although other mechanisms such as the maintenance of muscles at optimal force generating lengths also result from normal lordosis. In addition, the facets in contact have been suggested to bear significant load when lordosis is maintained by Adams and Hutton[13] and Shirazi-Adl et al[14].

At this point it would be interesting to observe the interplay between muscle and ligament forces and their loading effects on the lumbar joint during the performance of both squat style and stoop style lifts. Figure 4 shows the force-histories of some selected muscles; while Table 1 shows their instantaneous magnitude at the point in time when the peak extensor demand occurred – the most strenuous part of the lift.

The level of lordosis determines which tissue is ultimately responsible for supporting the low back moment during a given task. The active force generators (i.e. the musculature) are able to generate force when the spine is in any posture, albeit the force generating capability is modulated by many factors. On the other hand, the passive force generators (ligaments) only contribute force when passively stretched and such stretching occurs at the end range of joint motion. While contributions of individual ligaments over the full range of joint flexion has never been experimentally determined, analytical analyses have predicted that their primary role is to limit flexion and do not significantly impede flexion during moderate lumbar motion[15,16]. Returning to the example of subjects lifting reasonably demanding loads using a squat posture; it was acknowledged that they invariably elected to preserve enough lordosis as to not significantly recruit the ligaments, leaving the musculature with the responsibility to maintain equilibrium. Under these conditions, the ligaments and the disc in bending rarely contribute more than 4% of the restorative moment. In an effort to encourage ligament recruitment, work was conducted whereby subjects were requested to stoop rather than squat. Dropout of the extensor EMG signal confirmed that the load was supported by the ligaments; a phenomenon known as flexion-relaxation[17]. Workers are often observed in a stooped posture for lengthy periods of time but without a large load sustained in the hands. However, even in the stooped posture, subjects exhibit significant extensor EMG activity as soon as weight is taken in the hands. This evidence suggests that some mecha-

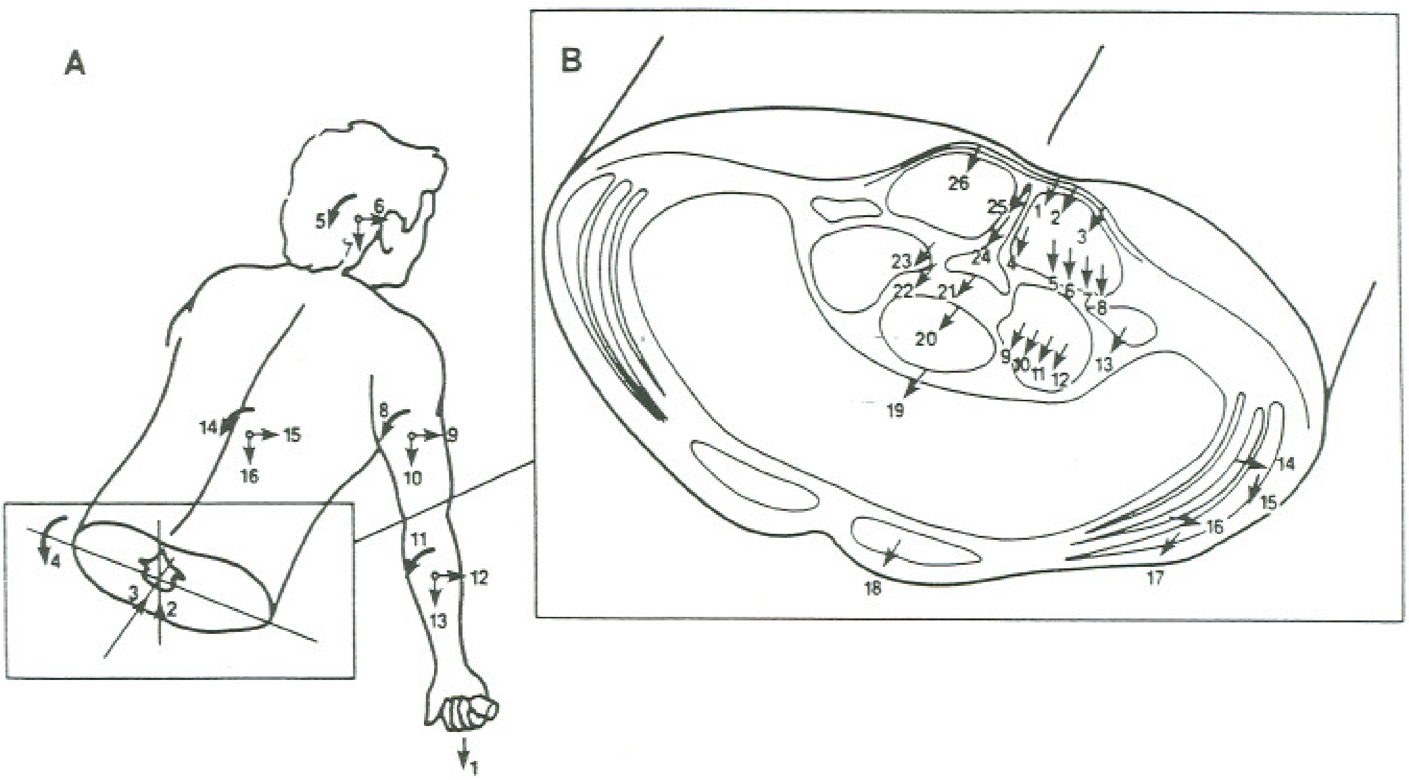


Figure 3 The reaction moment is determined from a rigid linked-segment model (moment 4, figure 5A) which is partitioned into restorative components provided by muscles (forces 1-18, figure 5B), ligaments (forces 19-26, figure 5B) and the disc in bending. Other forces, such as from IAP, were also evaluated. (Published with permission of Harper & Row, Philadelphia).

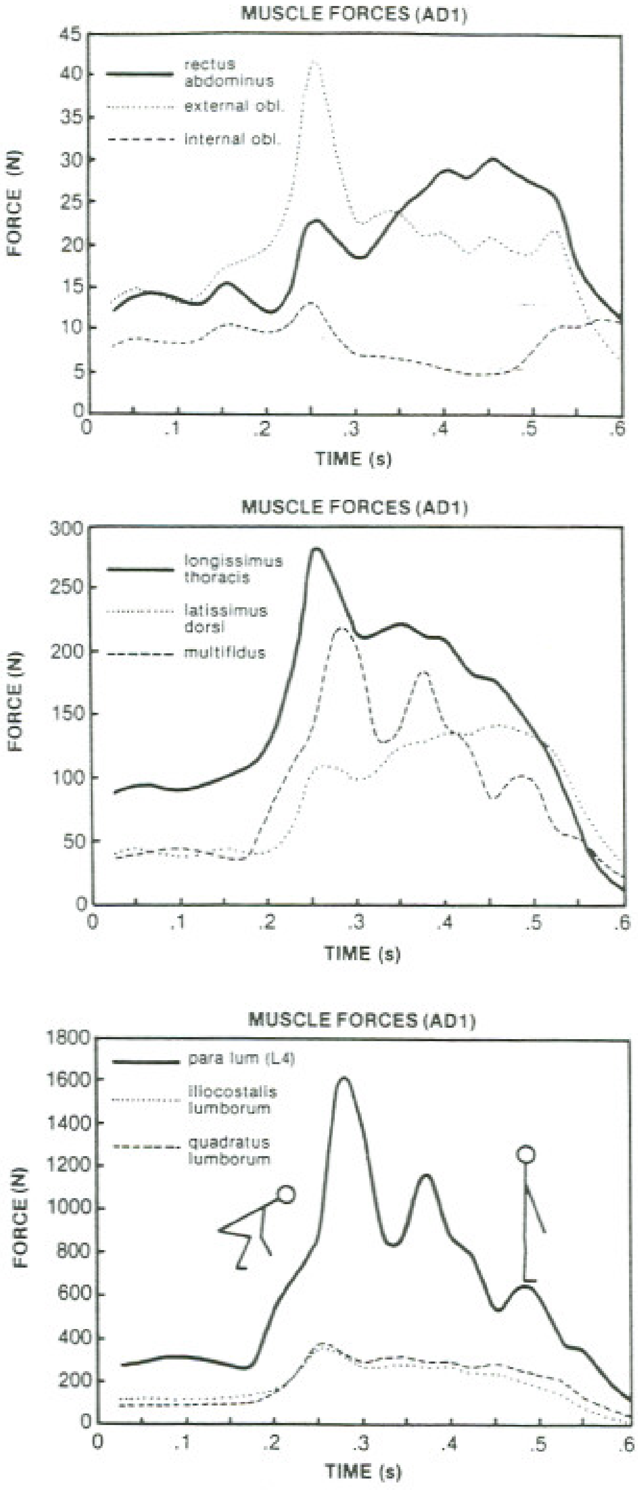


Figure 4 Individual muscle forces for a sample trial of a subject lifting 27 kg quite quickly in the sagittal plane. (Published with permission of Harper & Row, Philadelphia).

Table 1 Musculature Components for Reaction Moment Generation of 450 Nm at L4/L5 During the Point of Peak Loading for a Squat Lift of 27 kg

Muscle	Force (N)	Moment (Nm)	Compression (N)	Shear (N)
Rectus abdominis	25	−2	24	5
External oblique 1	45	1	39	24
External oblique 2	43	−2	30	31
Internal oblique 1	14	1	14	−2
Internal oblique 2	23	−1	17	−16
Longissimus thoracis pars lumborum L4	862	35	744	−436
Longissimus thoracis pars lumborum L3	1,514	93	1,422	−518
Longissimus thoracis pars lumborum L2	1,342	121	1,342	0
Longissimus thoracis pars lumborum L1	1,302	110	1,302	0
Iliocostalis lumborum pars thoracis	369	31	369	0
Longissimus thoracis pars thoracis	295	25	295	0
Quadratus lumborum	393	16	386	74
Latissimus dorsi L5	112	6	79	−2
Multifidus 1	136	8	134	18
Multifidus 2	226	8	189	124
Psoas L1	26	0	23	12
Psoas L2	28	0	27	8
Psoas L3	28	1	27	6
Psoas L4	28	1	27	5

NOTE: negative moments correspond to flexion while negative shear corresponds to L4 shearing posteriorly on L5.

nism is invoked to not allow the ligaments to support a strenuous moment without auxiliary muscular contribution. We can only speculate that this is an attempt to provide a system of neuromuscular control to protect the passive supporting tissues. Thus in most strenuous lifts, it appears that the burden of extensor moment generation during movement is relegated to the muscles.

Arguments that the musculature cannot supply the forces necessary to support the observed reaction extensor moments have been raised by Farfan[18]. The ability of a muscle to generate a moment is determined by the force and the perpendicular distance, or moment arm, to the fulcrum assumed to be the nucleus pulposus in this case. Our anatomic studies, and those of others show that the moment arms of many of the low back extensors can approach 10 cm in a healthy worker (McGill et al,[6]). In addition, Farfan[4] quoted muscle cross-sectional area data derived from a cadaver (Eycleshymer and Shoemaker,[19]). Several CT scan studies have demonstrated the much larger muscle areas of younger men who have not experienced the atrophy associated with inactivity of the elderly (eg. Reid and Costigan[5], McGill et al,[6]) (see table 2). If these areas are considered and credit is given to the full extent of the musculature that can contribute to lumbar extension, then the estimates of moments generated exclusively from muscular sources increase significantly. The muscular bulk of the thoracic components of iliocostalis lumborum and longissimus thoracis is not visible in a lumbar section but it must not be neglected in an estimate of the moment potential of the musculature. By assuming muscle can produce force at 35N/cm^2, then this author conservatively estimates the average total extensor moment potential of the extensor musculature to be over 500 Nm in healthy working men which is close to that experimentally measured by Troup and Chapman.[20]. Although, moments of this magnitude are rare in daily activity, it would be unfair to suggest that there *must* be other dominant sources of extensor moment when full credit has not been given to muscular sources.

Contributors to joint compression and shear

The components of muscular moment generation are detailed in Table 1 for the period of peak loading in a sample squat lift of 27 kg which produced a reaction moment in the low back of 450Nm. Knowledge of the individual muscle forces, subsequent joint moment and components of compression and shear that are imposed on the joint is quite useful information. It is the opinion of this author that compressive and especially shear components of muscular force have been greatly neglected during assessment of injury mechanisms. The very large magnitude of force in the pars lumborum laminae result from their large individual cross-sectional area. These forces produce a large proportion of the extensor moment. Negative moments observed in Table 1 correspond to the flexor contributions of abdominal co-contraction. The abdominal co-contraction in this lifting example, and most sagittal plane lifting tasks, was small at the instant of peak extensor moment.

The compression penalty from even mild abdominal activity can be observed from the table of individual muscle forces. To meet the requirement of the net moment, additional extensor activity is necessary to offset the flexor moment produced by the abdominals. However, this creates a double contribution to joint compression; compression from abdominal activity together with compression from the additional extensor forces needed to reach the required net moment. Even so, when all of the component forces are summed, the total predicted joint compression is less than what would have been predicted by the simple single equivalent muscle model introduced in an earlier section of this presentation. We have observed a varying ability of subjects to reduce compression on the spine for a given task that has ranged from 5 to 15% of that predicted by the 5 cm single equivalent model. In fact, in the complete absence of abdominal activity a single equivalent model with a moment arm of 7.5 cm often predicts comparable values of compression. Obviously, abdominal activity would result in a shorter equivalent moment arm. In most lifting cases, this reduction is large enough to reduce compression to subfailure levels of the vertebral endplate so that hypothesized compression reducing mechanisms (such as intraabdominal pressure) are no longer required. The ability of an individual to reduce compression appears to be determined by the degree of abdominal activity (the time course of muscle forces for a sample lift are shown in Figure 4). However, as is often observed in elite lifters, the abdominals are not completely silent and exhibit varying degrees of activity. This suggests that they are sacrificing minimum compression for some other as yet unknown benefit. Interviews with some elite lifters as to why co-contraction is observed often reveal that the lifters feel that it stiffens the trunk to prevent buckling of the spine. This idea will be addressed later in the discussion on intraabdominal pressure.

Negative shear forces from the muscles (shown in Table 1) correspond to L4 shearing posteriorly on L5. Hence, a very powerful anti-anterior shear mechanism is observed, in the tabulated forces, due to the obliquity of the pars lumborum extensors. These muscles help to offset the anterior reaction shear force from lifting a load when they are activated to presumably contribute extensor moment. These forces have been previously recognized to reduce load on the facet joints. Some subjects that we have tested have offset the reaction shear force almost completely, depending on the forward inclination of the disc (and trunk), and on the magnitude of force in these obliquely orientated pars lumborum fibres. This is further evidence to suggest that previous attempts to determine the joint load has resulted in serious overpredictions of both compressive and shearing components.

Table 2 Comparison of the muscle areas observed from planar CT Scans (through L4/L5) with those of other studies. Values are for both sides of the body. Note that most of the extensor bulk cannot be observed from scans at the L4/L5 level. Hence the extensor potential is greatly underestimated from the uncorrected area shown here.

	Muscle Areas of McGill et al (41)(cm^2)	Muscle Areas of Other Studies (cm^2)
Psoas	35.2	16.0 (1)
Erector Area (uncorrected)	45.1	31.7 (2)
Sacrospinalis	31.8	29.7 (1)
Multifidus	8.3	9.4 (1)
External Oblique	18.7	13.7 (3)
Internal Oblique	16.2	11.4 (3)
Rectus Abdominis	15.8	5.3 (3)
		10.5 (2)
		7.6 (1)
Transverse Abdominis	3.0	4.8 (3)
		7.5 (1)

SOURCES: 1. Farfan (4) – derived from cadaveric data of Eychleshymer and Shoemacher (19)
2. Reid and Costigan (5) – from CT scans of 16 male and 12 female living subjects.
3. Schumacher and Wolff (38) – from 21 cadavers

Evaluation of mechanisms hypothesized to reduce joint load

The motivation to postulate mechanisms that reduce joint compression came from the need to align load predictions produced by the simple single equivalent model (introduced earlier in this manuscript) with joint failure tolerance data. However, the anatomically detailed model outlined in this chapter has provided evidence pointing out the overestimation of loads which are predicted by the single equivalent model. In many cases, the more reasonable estimates of joint load from the detailed model negate the need for further reduction to reach tolerable levels. The following section strengthens this position with a review of the feasibility of each hypothesized compression reducing mechanism. Indeed, this process has assisted the evolution of knowledge pertaining to lumbar function.

Intraabdominal pressure (IAP):

The controversy over the relative benefits of the compression reducing tensile force of intraabdominal pressure acting over the diaphragm and pelvic floor, is well developed in the literature. Many have used intraabdominal pressure as a mechanism to reduce estimates of lumbar spine compression[21,22,23,24]. However, some have indicated that they believe the role of IAP in reducing spinal loads has been over emphasized[25,26,27]. In fact, some recent experimental evidence suggests that somehow, in the process of building up IAP, the net load on the spine is increased! Increased low back EMG activity with higher IAP was noted by Krag and co-workers[28] during voluntary valsalva manoeuvres. Nachemson and Morris[29] and more recently Nachemson et al[30] showed an increase in intradiscal pressure during a Valsalva manoeuvre indicating a net increase in spine compression with an increase in IAP, presumably a result of abdominal wall musculature activity. Nearly all models to date, that make use of IAP to reduce loads on the spine, neglect the compressive action of the necessary abdominal co-activity. Not only does compression result from direct abdominal activation but also the extensors must increase their efforts to offset the additional flexor moment generated by the abdominals further elevating the load penalty imposed on the spine.

Individual muscle force output from the model described in this chapter enables the evaluation of the net benefit/penalty of the buildup of IAP and concomitant abdominal activity. It is important to review briefly the pertinent components of the model for they have an important effect on force estimates. During formulation of the anatomical model, a diaphragm was incorporated into the rib cage that was scaled in accordance with the dimensions of a 50th percentile male. Its area and shape were confirmed from measures in the anatomy laboratory and from CT scans. Its normal surface area (presented to the plane of the L4/L5 disc) was 243 cm^2 and the centroid of this area was 3.8 cm. anterior to the centre of the T12 disc. It was difficult to justify the surface areas used by other models in the literature as they would not fit into our skeleton (compare with a 511 cm^2 pelvic floor used by Morris et al[31]; 465 cm^2 diaphragm used by Chaffin[21], also moment arms for diaphragm forces on the spine of 11.4 cm by Morris et al[31], 9.1 cm by Chaffin[21] and 7 cm by Troup et al[24]). The size of the cross-sectional area of the diaphragm and the moment arm used to estimate force and moment produced by IAP have a major effect on conclusions reached about the role of IAP.

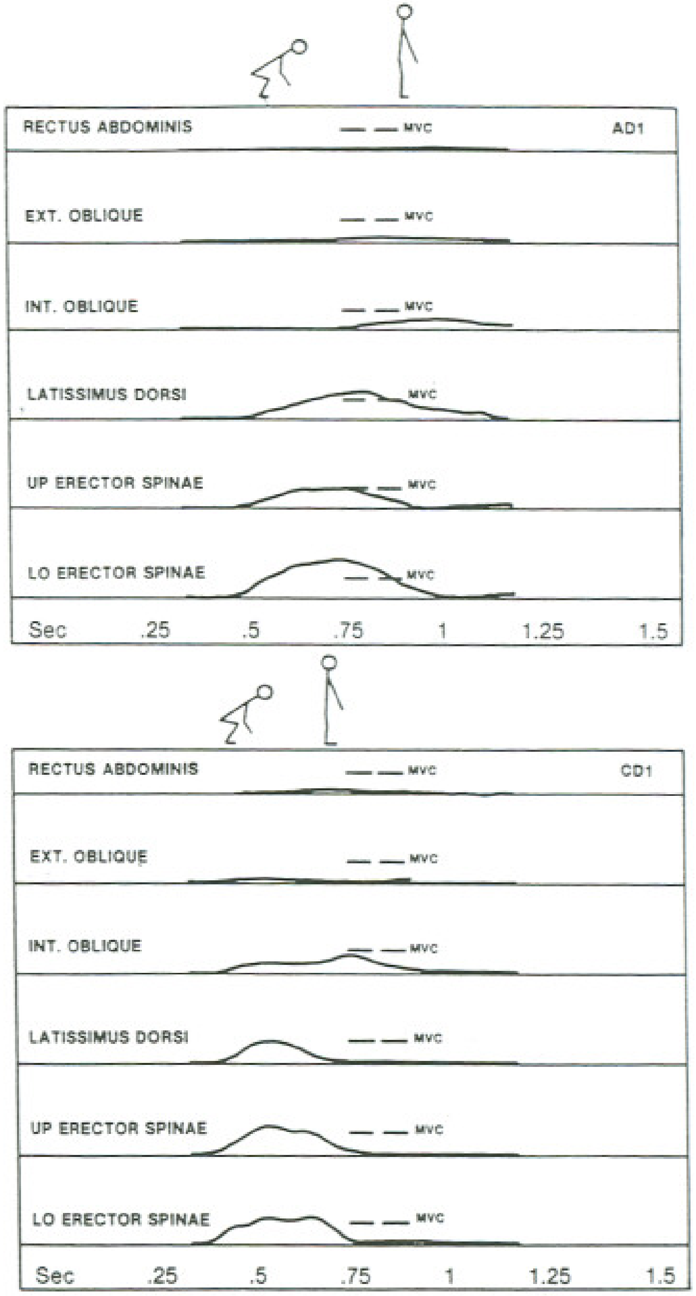


Figure 5 Records of linear envelope EMG from two subjects lifting 27 kg using a squat style.

It is usual to observe some level of abdominal activity during the lifting of loads. EMG records from two subjects are shown in figure 5. Note the relatively low activity in the abdominals during the period of maximum extensor moment generation while internal oblique activity increased at the end of the lift to apparently help balance the load in an upright standing posture. To evaluate the proposed effects of IAP the force-time histories of the individual abdominal muscles may be directly compared with the forces generated by IAP. During squat lifts, it appears that the net effect of the involvement of the abdominal musculature and IAP is to increase compression rather than alleviate joint load. (A detailed description and analysis of the forces are in McGill and Norman[2]). IAP was not measured directly in this first study but was predicted from various regressive strategies. Pressures were directly measured in a later study in conjunction with abdominal EMG which confirmed the magnitudes of the predicted pressure as well as observing pressure-time histories during a host of activities such as performing situps, running, lifting and jumping. During the lifts, the forces and moments created from IAP did not overcome the compression and flexor moment created by the abdominal activation necessary to create the pressure. The compression from abdominal activity versus compression relief from IAP is shown in Figure 6. This predicted finding agrees with experimental evidence of Krag et al[28] and Nachemson et al[29] which demonstrated increased intradiscal pressure with an increase in IAP. However, what is not lucid is the role of IAP given well documented evidence that appreciable pressures are generated during load handling tasks.

Farfan[4] has suggested that IAP creates a pressurized visceral cavity to maintain the hoop-like geometry of the abdominals.

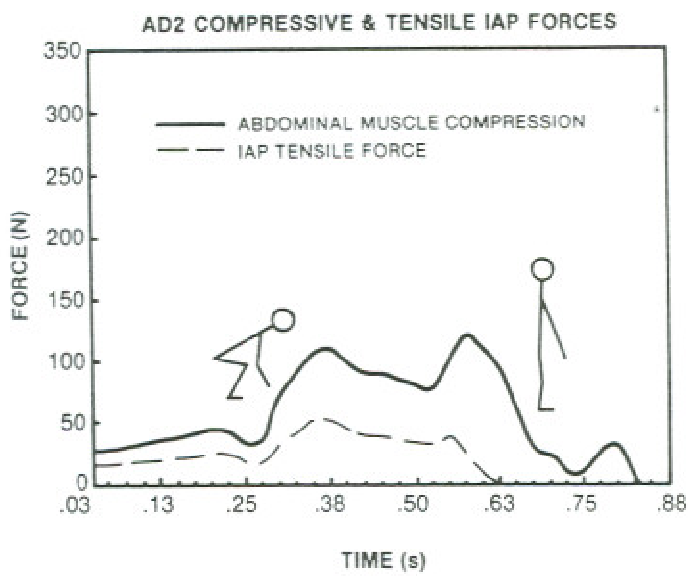


Figure 6 Compressive forces from the abdominal wall and tensile forces produced by IAP exerted on the diaphragm for a typical sagittal plane lift. (Published with permission of Taylor and Francis, London).

However, the compression penalty of abdominal activity cannot be discounted. In fact, the presence of abdominal activity is rather strong evidence that the mechanism of the lumbar spine does not work to minimize compression. Rather it appears that the spine prefers to sustain increased compression loads if intrinsic stability is increased. An unstabilized spine buckles under extremely low compressive load (eg. approximately 20N, Lucas and Bresler[32]). The geometry of the musculature suggests that individual components exert lateral and anterior-posterior forces on the spine which perhaps can be thought of as guy wires on a mast to prevent bending and compressive buckling. As well, activated abdominals create a rigid cylinder of the trunk resulting in a stiffer structure. Thus, a rigid trunk could prevent buckling and support shearing through the abdomen. This notion was echoed by a national class weight lifter after recently lifting 264 kg in our laboratory.

In a related study, we attempted to quantify the affects of an abdominal belt on lumbar function. Measures of IAP, and abdominal and erector spinae (adjacent to L3) activity were collected during squat lifting while both wearing and not wearing an abdominal belt. For repeated trials of identical lifts, wearing the belt significantly increased IAP while erector spinae activity remained unchanged. However, an unexpected finding was that holding the breath significantly increased IAP and tended to reduce erector spinae activity although this reduction was not significant. While this may be an indication of extensor assistance from breath holding it is not conclusive evidence that compressive loads on the spine are reduced.

There is no doubt that increased IAP is commonly observed during many activities as well as in those people experiencing back pain. However, the experimental evidence of others and the explanation of forces presented here still does not provide an unequivocal explanation. Work is continuing in our laboratory, and in others, in an effort to explain the presence of increased IAP during exertion.

Muscular activation of the lumbodorsal fascia (LDF)

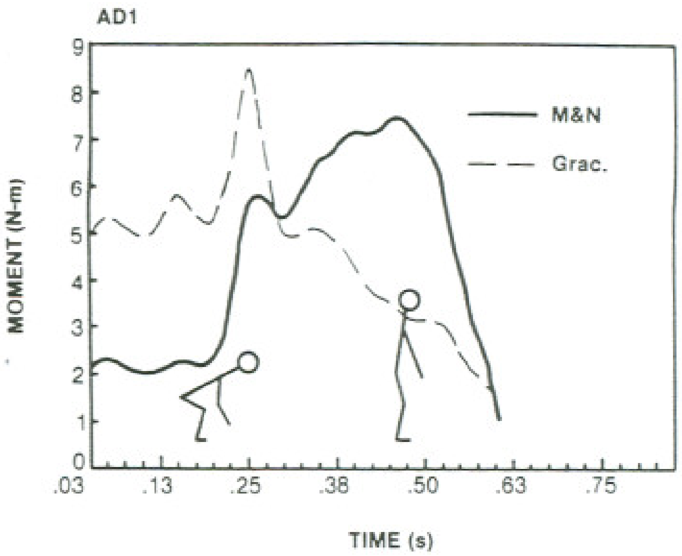
Recent studies have attributed various mechanical roles to the LDF. Suggestions have been made (Gracovetsky et al[11]) that lateral forces generated by internal oblique and transverse abdominis are transmitted to the LDF via their attachments to the lateral border. This lateral tension was hypothesized to increase longitudinal tension, from Poisson's effect, pulling in the direction of the posterior midline of the lumbar spine causing the posterior spinous processes to move together resulting in lumbar extension. This proposed sequence of events forms an attractive proposition because the LDF has the largest moment arm of all extensor tissues. As a result, any extensor forces within the LDF would impose the smallest compressive penalty to vertebral components of the spine. The problem with this hypothesis is that abdominal forces have never been experimentally proven to contribute to extension, nor can this author agree with the mathematical representation of the involved tissues (presented by Gracovetsky et al[11]).

The assumption that the muscular source of LDF extensor moment is solely from the abdominals is questionable. In fact, lateral tension applied to the LDF of a cadaver in our anatomy laboratory did not appear to cause visible spine extension because the force seemed to be interrupted by anchor points on the ilium. These attachment points to the posterior iliac spine are very rigid and marked by the "dimples of venus". Review of the anatomic literature (eg. Bogduk and Macintosh[33]) and cadaveric dissection provided evidence that latissimus dorsi was in a position to increase tension in the LDF and thus transmit extensor forces to the low back. An investigation was conducted to evaluate the potential of the LDF to generate extensor moment by 2 methods: one to observe the effects of the LDF activated by the abdominals as proposed by Gracovetsky et al[11] and another to study the possible involvement of latissimus dorsi in generating tension within the LDF (McGill and Norman[4]). The optimization strategy used by Gracovetsky et al[11] allowed the abdominals to generate large forces during extension. However, our EMG records could not confirm this prediction of large abdominal activity so that much lower abdominal forces resulted from the model (refer to figure 5). Thus the moment contribution from the abdominal method was quite low and never more than 5% of the required reaction moment demanded for dynamic equilibrium (see figure 7). However, the method of generating moments from the latissimus dorsi produced similarly low moments although the extensor contribution from latissimus dorsi occurred at the same time as the extensor demand as can be noted by the burst of latissimus dorsi activity in phase with that of the primary extensors (figure 5). On the other hand, the abdominal activity did not always correspond to the point in time where extension moment was required, which makes the abdominal strategy even harder to rationalize as an extensor mechanism. Regardless of the choice of LDF activation strategy, the LDF contribution to the restorative extension moment was negligible compared with the much larger low back reaction moment required to support the load in the hands.

Although the LDF does not appear to be a significant active extensor of the spine it is a strong tissue with a well developed lattice of collagen fibres. Its function may be that of an extensor muscle retinaculum (Bogduk and Macintosh,[33]). The tendons of longissimus thoracis and iliocostalis lumborum pass under the LDF to their sacral and ilium attachments. Perhaps the LDF provides a form of "strapping" for the low back musculature. Recently, a new avenue of work was undertaken by Tesh et al[34] who suggested the LDF may be more important for supporting lateral bending. No doubt, this insight will be pursued in the future.

Clinical implications of tissue loads

Undoubtedly, the patterns and conditions of tissue failure, fatigue and creep depend on the time course and magnitude of the applied load. Thus, knowledge of individual tissue forces and loads is essential for evaluation and analysis of clinical



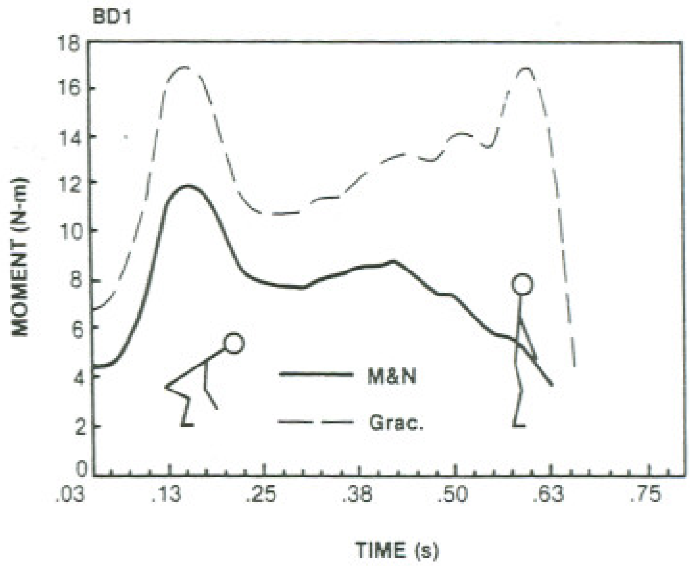


Figure 7 The LDF moment contribution calculated by the McGill and Norman method (lat. dorsi activation) and Gracovetsky method (abdominal activation) for two different subjects (AD1 and BD1). The contribution of the LDF is mall regardless of the activation strategy. The peak extensor moments generated during these lifts were 450 and 306 Nm for subjects AD1 and BD1 respectively. (published with permission of Butterworth, England).

disorders that are "mechanical" in origin.

The work that we have reported over recent years has demonstrated the extraordinary magnitudes of forces within the various components of the trunk musculature, even during non-strenuous tasks. Several explanations for the underestimation of muscular loads in the early biomechanical studies were provided previously in this paper. While these forces have been interpreted for their mechanical role, clinicians have expressed interest in their potential to cause injury. Certainly, one could implicate large muscle forces in any soft tissue injury. However, damage to bony attachments remain a possibility which perhaps has been wrongfully attributed to alternate mechanisms. One such example follows.

Pain in the sacro-iliac region is common and often attributed to disorders of the SI joint itself or the iliolumbar ligament (eg. Dontigny[35], Resnick et al[36]). For this reason the role of the musculature has been neglected. Earlier in this communication, it was shown that a large proportion of the extensor musculature obtains its origin in the SI and posterior superior iliac spine (PSIS) region by the excellent anatomical work of Macintosh and Bogduk[8]. The area of tendon-periosteum attachment and extensor aponeurosis is relatively small in relation to the volume of muscle in series with the tendon complex. From this, a hypothesis evolved that the seeming mismatch of large muscle tissue to small attachment area for connective tissue places the connective tissue at high risk of sustaining micro failure, resulting in pain (McGill[37]). Knowledge of the collective muscle force-time histories enables speculation about one-time failure loads and cumulative trauma. For example, if the forces of muscles that originate in the SI region are tallied for the trial illustrated in Table 1, then the total force transmitted to the SI region during peak load exceeded 6.5 kN. Such a load would lift a small car off the ground!

The failure tolerance of these connective tissues is not known, which make speculation over the potential for micro-failure difficult. No doubt, the risk of damage must increase with the extremely large loads observed in the extensor musculature and with the frequency of application. Task analysis of many industrial tasks show lifting 3 containers per minute in excess of 18 kg. over an 8 hr. day is not unusual, suggesting the potential for cumulative trauma is significant.

This mechanical explanation may account for local tenderness on palpation associated with the majority of SI syndrome cases. As well, muscle strain and spasm often accompanies SI pain. Nonetheless, treatment is often directed toward the articular joint despite the extreme difficulty in diagnosing the joint as the primary source of pain. While reduction of spasm through conventional techniques would reduce the sustained load on the damaged connective fibres, patients should be counselled on techniques to reduce internal muscle loads through effective lifting mechanics. This was a single example, of which there may be several, where knowledge of individual muscle force-time histories suggested a mechanism of injury for which a specific treatment modality would be prescribed.

Future directions for research related to Chiropractic

In the introduction of this review, a view was presented that further development of chiropractic hinged on public acceptance of the profession. The remaining impediment to widespread acceptance was a lack of scientific description of various physical techniques and manipulations. We are all aware of the difficulties in such research. However, recent efforts in the research community have begun to examine the mechanics of torsion in the lumbar tissues which is extremely relevant to Chiropractic given the dominance of rotational manoeuvres during manipulation. Our most recent work has been the improvement of a 3 dimensional spine model from which tissue loads during rotation can be estimated. At this most preliminary stage, it appears that the ligaments cannot be loaded in torsion unless the spine is flexed leaving the responsibility to resist torsion to the disc and facet complex. Nonetheless, clinical techniques that first flex the spine prior to axial torques are directly loading the ligaments.

This type of modelling approach provides the load histories within individual tissues and appears to be the only feasible way to formulate a mechanical description of a manoeuvre such as chiropractic manipulation. In fact, while chiropractic is perhaps the "purest" form of treatment that relies on the loading of specific tissues, there is some speculation that tissue strain invokes a biochemically based chain of events to promote healing. However, the link to understanding the healing process begins with the ability to assess internal tissue loads. This research is well underway, and the public perception of the chiropractic community in the future will depend, in no small amount, on the ability to provide mechanically based rationale, and explanation of specific chiropractic treatment.

Acknowledgement

Early studies reported in this review were supported financially by the Defence and Civil Institute of Environmental Medicine, Toronto and recent work by the Natural Sciences and Engineering Research Council of Canada. The author also wishes to acknowledge the collaboration with colleagues in the Department of Kinesiology, University of Waterloo, and especially Dr. R.W. Norman, who contributed to work reported throughout this review.

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