Biomechanical Basis for Manual Lifting Guidelines

Arun Garg, Ph.D.

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CHAPTER 3

BASIS FOR GUIDE: BIOMECHANICAL APPROACH

Arun Garg

Biomechanics deals with the laws of physics and engineering concepts applied to the human body to determine motions of body segments and the forces acting on various parts of the body (muscles, ligaments, joints, etc.). It is an attempt to combine physics and engineering with anatomy and physiology. Thus, biomechanics is a multidisciplinary activity. Occupational biomechanics is an applied discipline in the general field of biomechanics. According to Chaffin and Andersson (1984), "occupational biomechanics can be defined as the study of the physical interaction of workers with their tools, machines, and materials so as to enhance the worker's performance while minimizing the risk of future musculoskeletal disorders."

Since back pain has a multifactorial etiology, its prevention requires a biomechanical approach which is equally broad. In addition to engineering and life sciences, occupational biomechanics must include epidemiology, behavioral sciences and psychophysical factors (Troup, 1979). Occupational biomechanics deals with those human disorders and performance limitations which are either produced or aggravated by the mismatch between the human physical capacities and human performance requirements (job requirements) (Chaffin and Andersson, 1984).

Occupational biomechanics is extremely useful in (i) quantifying stresses on various body parts (musculoskeletal system) from manual handling of loads, (ii) determining the risk of an injury from these stresses, especially when combined with epidemiological studies, and (iii) evaluating alternative designs and methods to prevent potential injuries. Knowledge about mechanical and physiological functions of the spine is useful in determining: (i) etiologies of low-back pain syndromes, (ii) quantifying stresses on the spine due to maintenance of body posture and external forces, and (iii) methods to prevent potential back injuries. Additionally, biomechanics plays an important role in determining treatment and care programs for the already injured. This chapter concentrates on the biomechanics of the low back as a basis for a load handling limit.

OCCUPATIONAL BIOMECHANICS OF LOAD LIFTING

It is a well established fact that the stresses induced at the low back during manual materials handling are due to a combination of the weight lifted and the person's method of handling the load. The internal reaction forces needed to equilibrate the body segment weights and external forces (such as weight of the load being lifted) are supplied by muscle contractions, ligaments and body joints. More importantly, the human body acts as a lever system. Specifically, the external forces applied with the hands as well as the body segment weights create rotational moments or torques at various body joints. The skeletal muscles are positioned to exert forces at those joints in such a manner that they counteract the moments due to the load and body segment weights. Since the moment arms of the muscles (and ligaments) are much smaller than the moment arms of the external forces and body segment weights, small external forces produce large muscle, ligament and joint forces. However, the advantage is that the muscles can produce large motions with small degrees of shortening.

For example, consider the task of holding a load, L_h, in both hands with the forearm-hand segments held horizontally and the upper arms held vertically (Figure 3.1). Under static equilibrium conditions:

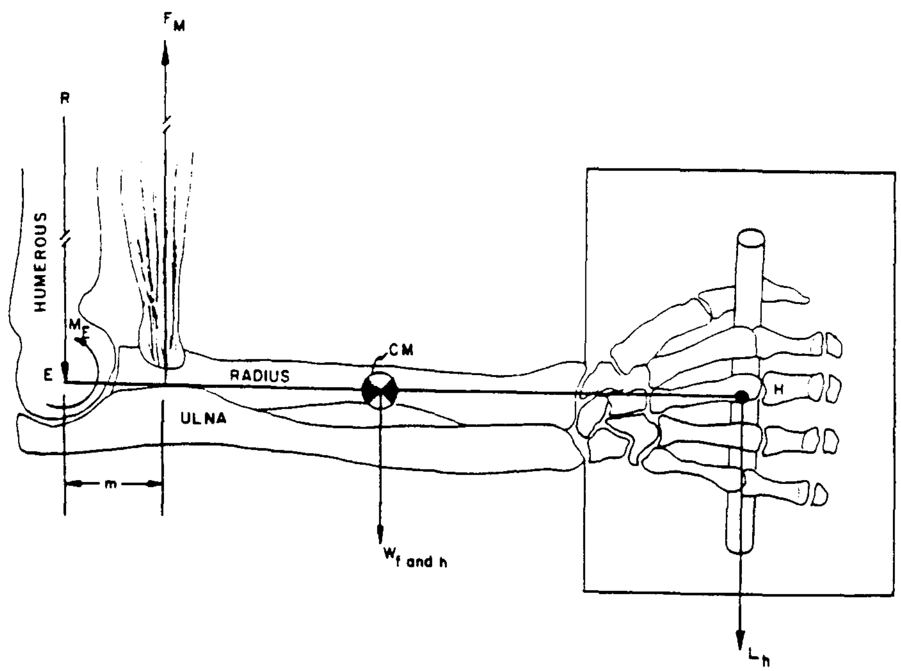


Figure 3.1: Analysis of internal muscle force F_M and joint reaction force R created by weight of forearm-hand $W_{f\&h}$ and load L_h held horizontal. (Chaffin and Andersson, 1984)

$$M_E = 0$$

$$- M_{load} - M_W + M_m = 0$$

where M_{load}, M_w and M_m are moments at the elbow joint due to weight of the load, weight of the lower arm and muscle force. M_E is the net moment at the elbow joint. If it is assumed that the primary muscle action is provided by the biceps brachii and its moment arm is 5 cm from the elbow center of rotation, then for an average male holding a 10 kg mass (98 N load) in both hands (49 N load in each hand):

$$35.5 \text{ cm } (-\ 49 \text{ N}) + 17.2 \text{ cm } (-\ 15.8 \text{ N}) + 5 \text{ cm } (F_M) = 0$$

$$F_M = 402 \text{ N}$$

Thus, holding a load of 49 N (5 kg mass) in each hand creates an internal muscle force requirement about eight times greater (402 N). This mechanical disadvantage of muscles due to their small moment arms can produce high stresses in muscles in certain body postures, even when lifting relatively light loads.

The elbow joint reactive force, R_E, acting on the distal trochlea of the humerus can be determined by balancing the external and internal forces along the vertical axis:

$$F_E = 0$$

$$R_E + F_m - W_{f\&h} - L_h = 0$$

$$R_E + 402 - 15.8 - 49 = 0$$

$$R_E = -\ 337 \text{ (downward)}$$

where $W_{f\&h}$, L_h and F_E are weight of the forearm and hand, weight of the load, and net reactive force at the elbow joint in the vertical direction.

This shows that the high forces created by muscles during a voluntary exertion produce equally large forces on adjoining skeletal structures. This concept of muscles loading skeletal structures is extremely important in the biomechanics of low-back injuries as handling of light loads in certain postures can create large compressive loads on the lumbar spine structures.

One of the assumptions in the above analysis of internal forces is that a single muscular action accounts for the internal force. Often, several muscles share the moment requirement at a joint. For example, if two muscles flex the elbow--the brachialis and the biceps brachii--then there are four unknowns (F_{m1}, F_{m2}, $_3$ and R) and three independent equations ($M_E = 0$, $F_x = 0$, and $F_y = 0$) in Figure 3.2.

Thus, a unique solution is not possible without certain assumptions, as the system is statically indeterminate. Unfortunately, it is not clear which particular biomechanical objective is dominant during the performance of a manual task. Further, subtle changes in body posture may shift load from one muscle group to another and there are wide variations in individual motor behavior patterns. Details of this can be found in Chaffin and Andersson (1984).

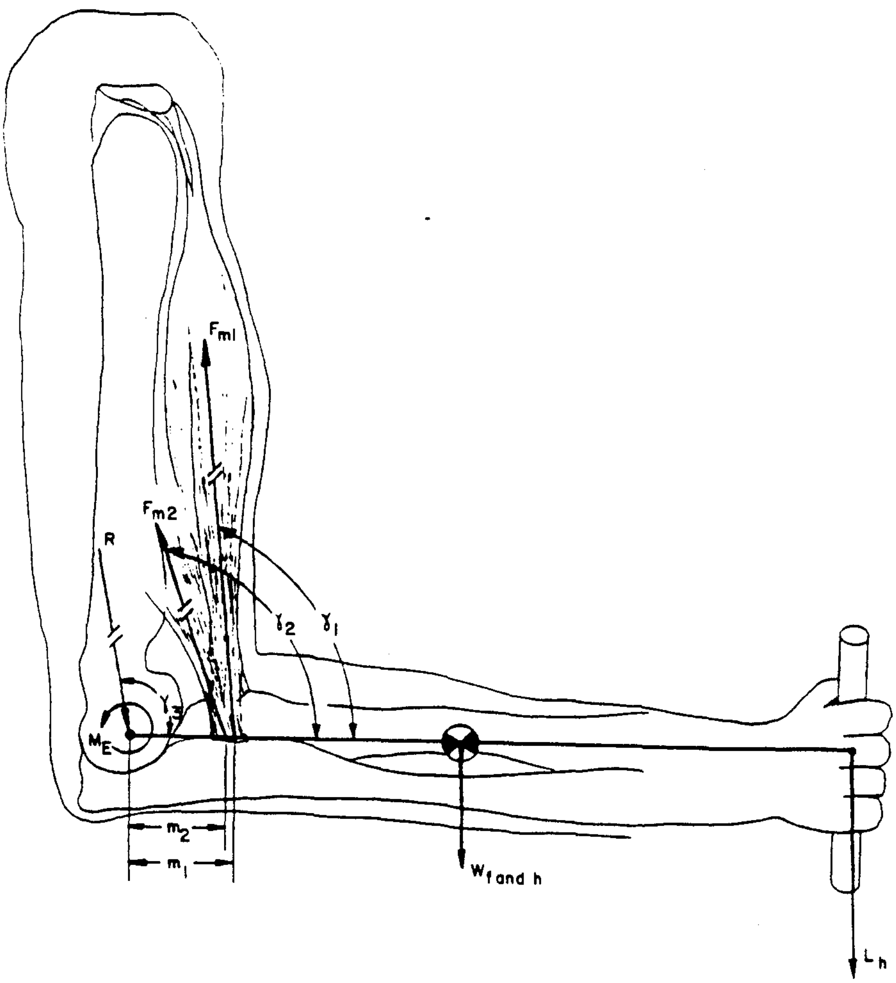


Figure 3.2: Two-muscle system for elbow flexion. (Chaffin and Andersson, 1984)

The second major assumption in the model of the elbow was that there was no antagonistic activity, i.e., elbow extensors were silent. When analyzing the lower back this may be a reasonable assumption when there is no lateral bending and/or axial rotation of the spine involved (Bean et al., 1988). Several studies have shown the presence of antagonistic activity, especially when the trunk is laterally bent and/or axially rotated (Portnoy and Morin, 1956; Andersson et al., 1977; Ortengren and Andersson, 1977; Zetterberg et al. 1987; Pope et al., 1987; Schultz et al., 1987). Biomechanically, the presence of antagonistic muscle activity should result in greater compressive force on the spine.

Lastly, it was assumed in the elbow example that all restorative moment was provided by the muscles. It appears that under certain load lifting conditions, such as for some strenuous tasks, the ligamentous tissues of the lumbar spine may passively provide significant resistance (restorative moment) to external moments (Gracovetsky et al., 1981: Gracovetsky et al., 1985; Anderson et al., 1985; Schultz et al., 1987). This would reduce the muscle contraction forces required to develop necessary restorative moment. At present, it is not clear under what conditions antagonistic muscles and ligaments play an important role, the extent of their involvement and the biomechanical objective function that the body attempts to minimize.

BIOMECHANICAL BASIS FOR BACK INJURY

Injury of the lumbar spine occurs as the limits of maximum stress or strain of the tissues are exceeded. This may happen due to a single strenuous effort or due to repeated loadings. In one case a truly accidental event leads to trauma. More often it is a situation in which there is no disruption in the normal pattern of work, but only an unexpected onset of pain during an everyday task. In the later case it might be assumed that local degenerative changes have increased the susceptibility to a sudden onset of symptoms.

One of the difficulties in analyzing the cause of a back injury is that it can take place without pain because neither the facets of the apophyseal joints (synovial membrane has nerve endings) nor the intervertebral discs receive a nerve supply (Troup, 1979). Thus, two major load bearing tissues of the spine can be injured without pain. "Evidence of old, stable fractures is common in people with no history of injury and there are innumerable cases in which the onset of pain was delayed for 24 hours or more after the injury." (Troup, 1979). If the intervertebral discs and/or apophyseal joints are repeatedly injured, degenerative changes may set in; though radiographic evidence of degeneration is equally prevalent in those with or without low-back pain (Pope et al., 1984). It is generally believed that the low-back pain from occupational and non-occupational physical stresses may be the culmination of a series of truly accidental, but at the time painless, injuries (Troup, 1979). Irreversible mechanical derangement of the intervertebral joint (for example, a decrease in disc height, changes in relative position of the apophyseal joints) may produce clinical symptoms long after the acute phase of injury (Brinckmann et al., 1988). To what extent the intervertebral disc is a source of pathology is not clear, but disc degeneration is often linked to low-back disorders. It appears that chronic low-back pain is often the result of degeneration of the disc (Morris et al., 1961; Armstrong, 1965; Rowe, 1971).

The compressive strength of the lumbar vertebral bodies and intervertebral discs has been the focus of many studies because of clinical interest in disc diseases and their causes. In the healthy spine, the disc is stronger than the vertebral bodies (Roaf, 1960). Cadaver studies show that the cartilage end plates that distribute the compression loads to the bodies of vertebral segments fail (Roaf, 1960; Armstrong, 1965; Hutton and Adams, 1982). A person can sustain a compression injury without a radiographic visible fracture. End-plate fracture may result in intracorporeal nuclear protrusion and loss of disc turgor (Roaf, 1960). Loss of nuclear turgor can cause abnormal mobility between the vertebral bodies and the disc is now vulnerable. Later a relatively minor injury (such as compression or flexion movement) can cause a severe and incapacitating annular prolapse (Roaf, 1960).

The large variations in strength of the cadaver columns may indicate that the cartilage end plates of some people were already weakened by prior stresses, with resulting microfractures and scarring. If true, this would contribute to the disc degeneration that now is acknowledged as being necessary before the more common and most serious discogenic low-back problems can develop. In other words, evidence indicates that large spinal-disc compression forces can be produced by muscular exertion, especially when lifting (Chaffin and Andersson, 1984). These repeated compressive stresses can be sufficient to cause microfractures in the cartilage end plates and subchondral bone of the vertebral bodies. It is believed that repeated microfractures and bone scarring of the cartilage end plate leads to a weakening of the annulus fibrosus of the disc, which thus protrudes into the spinal canal, as shown in extreme form in Figure 3.3 (Chaffin and Anderson, 1984). End plate fractures are always accompanied by an irreversible decrease in disc height and a permanent increase of the radial disc bulge (Brinckmann and Horst, 1985). Bulging or rupture of annulus fibrosus can cause nerve root compression or distortion of the ligaments around the disc or at the posterior facet joints.

It is believed by Rowe (1969) that 70-80% of all chronic low-back pain will be diagnosed as discogenic after a period of repeated episodes. At the very least, degeneration and the narrowing of the disc that results from it will contribute to a more unstable spinal structure. In this regard, Fiorini and McCammond (1976) state: "Many damaged discs showing radiographical narrowing may not cause discogenic back pain at all if there is no disc herniation, but may cause facetogenic back pain by causing subluxation and malapposition of the interarticular joints."

Some evidence that disc degeneration is accelerated by physical stresses has been developed by Hult (1954). He reported that narrowing and osteophyte developments of the discs and adjacent vertebral bodies was 1-1/2 times greater in those people engaged in heavy physical labor than in sedentary workers. In summary, although several low-back injury mechanisms have been proposed (Farfan et al., 1970), disc compression has been named as an important measure of low-back stress causing end plate fracture, disc herniation and nerve irritation.

The above disc degeneration theory implies that assigning a cause for low-back pain cannot be based simply on the immediate circumstances at the time when the pain first developed. In fact, most low-back episodes do not suddenly start with a "jabbing pain," although these cases are easily remembered and reported. Rather the symptoms more often are slow to develop, with stiffness, dull aching pain and, finally, incapacitating discomfort, which occurs possibly hours or even days later.

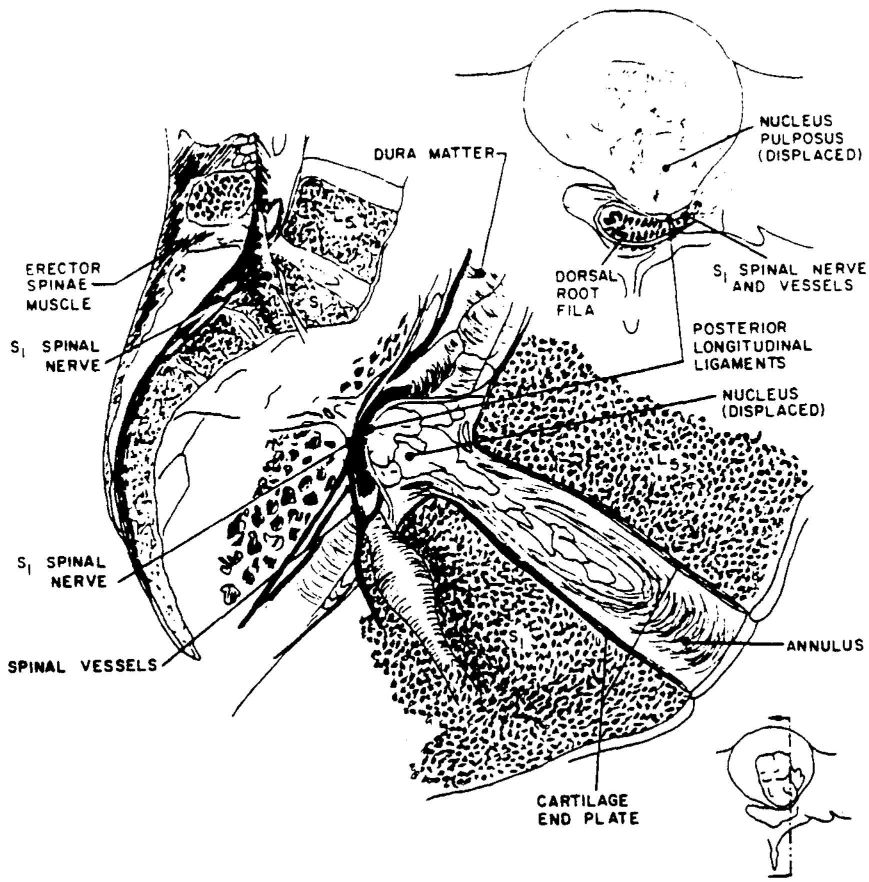


Figure 3.3: Example of L_5/S_1 disc herniation resulting in spinal nerve root compression. (Chaffin and Andersson, 1984)

The disc shear forces, experienced in flexion, extension, lateral flexion and axial rotation, also need some attention. These shear forces are resisted primarily by the posterior facet joints in the lumbar spine and the annulus fibrosus of the disc. Under normal physiological conditions, the facets can resist shear forces (Fiorini and McCammond, 1976). However, the limit to such forces is not well documented (Schultz et al., 1979). Cyron et al. (as stated in Troup, 1977), Lamy et al. (1975) and Farfan et al. (1976) reported about 2,000 to 3,000 N of shear forces as the failure limits for the articular facets. If a disc space is narrowed by degeneration, it may result in abnormally high stresses on the facet joints (Fiorini and McCammond, 1976). These authors also suggest that interarticular facets, being unstable, are more vulnerable to injury and, with their rich blood supply and nerve endings (in synovial membrane), are likely to cause considerable pain. Further, facet joints may be injured simultaneously whenever there is a disc injury or may be injured exclusively without a disc injury.

Virgin (1951), Markolf and Morris (1974), Gracovetsky and Farfan (1986) and Gracovetsky (1986) proposed that instead of intradiscal pressure (nucleus), the annulus is the main load transmission structure. Further, the load transmission capacities of the disc are time dependent because the annulus is made of viscoelastic collagen fibers. Therefore, the amount of load that may be safely lifted depends on time, i.e., heavy loads must be lifted quickly because the amount of load that the annulus can transmit decreases with time. However, Anderson et al. (1985) concluded that annulus rupture due directly to strain on the tissue appeared unlikely, though it could be a problem indirectly via disc degeneration. Hickey and Hukins (1980) concluded that compression was likely to cause end-plate failure rather than the annular failure. According to Roaf (1960), the pressure is mainly transmitted through the annulus only if the nucleus is no longer fluid (for example, older subjects). Roaf (1960) reported that compression of specimens from older subjects in which the nucleus was no longer fluid resulted in either tearing of annulus, general collapse of the vertebrae or a marginal plateau fracture. Nachemson and Morris (1964) suggested that high intradiscal pressure could cause high tangential stress (tensile forces) in the posterior part of the annulus. Thus, high intradiscal pressure could contribute to low-back pain by rupture of the fibers of the annulus, especially in those whom this structure is weakened. Also, a degenerated disc would considerably increase the vertical load on the annulus.

Some researchers (Farfan et al., 1970; Hickey and Hukins, 1980; Farfan and Gracovetsky, 1984; Farfan, 1985; Gracovetsky and Farfan, 1986; Gracovetsky, 1986) believe that torsion injuries are more frequent and more damaging than compression injuries. Torsion injuries involve peripheral damage to both the disc and the facet joints. These and other issues are discussed later under biomechanical models and asymmetric lifting.

The following studies and others mentioned in the next subsection provide additional support for using compressive force as one of the criteria in developing "safe" load limits.

Krämer and Gritz (1980) reported that body stature underwent a normal dirunal shrinkage of the order of 1.1% attributable to loss of disc height due to the cumulative effects of loading during the day. Fitzgerald (1972) demonstrated losses in overall stature in healthy young adult males with the shoulder load of 88 N (erect posture) for 20 minutes and recovery took about 10 minutes. Eklund and Corlett (1984) reported that shrinkage rates were closely related to the calculated levels of spinal compression at the level of L_3.

Whenever a compressive load produces a tissue pressure which is greater than the osmotic pressure, fluid is expelled from the disc (called creep effect, (Kazarian, 1975)). When the fluid is expelled from the disc, it narrows and this affects the dynamics of the intervertebral joint. The area of contact between the bearing surfaces of the apophyseal joints changes as these are compressed and strained and the intervertebral joint becomes stiffer. Thus, under the load, the capacity of the tissues to dissipate energy and their ultimate strength are reduced (Kazarian, 1975).

COMPRESSIVE STRENGTH OF LUMBAR SPINAL COLUMN

The maximal amount of compression that can be tolerated by the lumbar spinal column has been estimated from axial compression loading tests on cadaver columns. Load to failure performed in a single test determines the ultimate strength of a structure. The ultimate compressive strength of human lumbar vertebrae is presented in Figure 3.4. In general, the ultimate compressive strength varies between approximately 3 kN and 12 kN (Jäger, 1987). The mean value of lumbar compressive strength from the 283 experiments shown in Figure 3.4 was calculated as 5.1 ± 2.3 kN (Jäger, 1987). Bartelink (1957) found that discs were destroyed by forces ranging from 1.56 kN to 6.68 kN with a mean of 3.12 kN. Recently, Brinckmann et al. (1988) data showed ultimate compressive strength ranging from 2.1 kN to 8.8 kN (with the exception of one cadaver column). Also, an analysis of their data showed that about 30% of cadaver spinal columns fractured at loads below 4 kN and about 63% of cadaver spines had the ultimate compressive strength of 6 kN or less. Large variability in measured compressive strength within a study and between different studies makes it difficult to provide a single value of spinal compressive strength that would be valid for all persons.

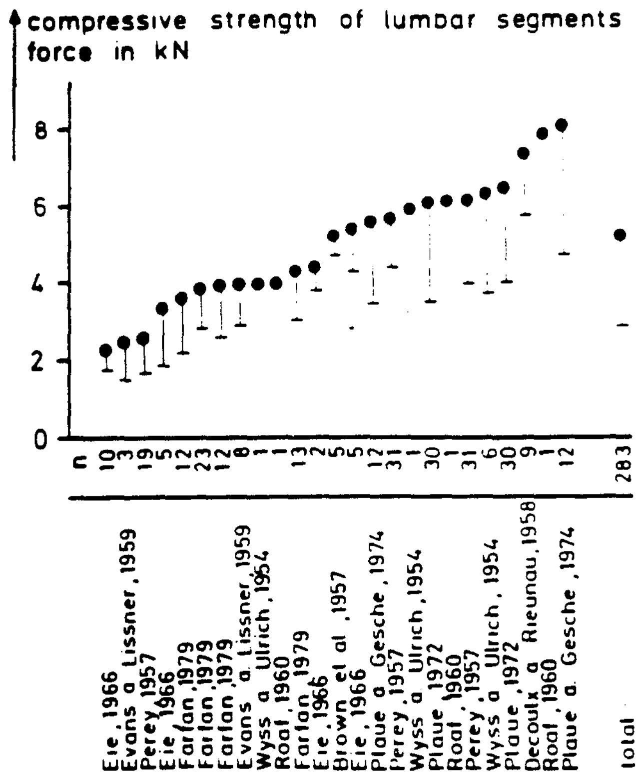


Figure 3.4: Morphological studies to determine the fracture load of lumbar segments of various length. (Jäger, 1987)

The strength of vertebral bone is primarily related to its mineral content (Hansson, 1977), bone material properties (Brinckmann et al., 1988), bone dimensions and experimental techniques (Brinckmann et al., 1988). Compressive strength of vertebral bodies varies with duration of applied stress (Perey, 1957), i.e., the longer the spine is subjected to a given compressive load, the greater the likelihood of damage. It also depends upon strain rate (Hutton et al., 1979; Troup and Edwards, 1985), i.e., how quickly the loading is applied. Hutton et al. (1979) reported a significant increase in breaking load at the faster strain rate due to viscoelastic properties of the bone. It also depends upon existing degenerative changes (Hansson and Roos, 1981). When degenerative changes have set in the spine becomes more susceptible to non-bony injury (Troup, 1979).

The disc is the strongest in young adults and weakens slowly with age (Perey, 1957; Hutton and Adams, 1982). Figure 3.5 illustrates data from Evans (1959) and Sonoda (1962) by age group. In general, the data from male cadavers disclose a mean of about 6.6 kN (675 kg) for those under 40 years of age and 2.4 kN (250 kg) for those over 60 years of age before the cartilage end plates begin to exhibit microfractures.

Sonoda (1962) estimates that the female's spinal compression tolerance is about 17% less than the male's. This would be consistent with the smaller force-bearing area of the vertebral bodies in a woman's spine.

The vertebrae are also subject to fatigue failure owing to repeated loading (Troup, 1977; Brinckmann et al., 1988). Under repeated loading by a compressive force, the height of a motion segment specimen (two vertebrae with intervening disc) decreases slowly due to viscoelastic deformation and water loss of the disc. A sudden irreversible height loss is interpreted as a compressive fracture (fatigue fracture). Brinckmann et al., (1988) showed that the probability of encountering a fatigue fracture increased both with the number of load cycles and the magnitude of relative cycle load (% of ultimate compressive strength). "At loads below 30% of compressive strength, practically no fractures were observed up to 5000 load cycles. At loads above 70%, specimens will sustain only a few cycles to fracture," (Brinckmann et al., 1988). Results of this study are presented in Figure 3.6. It was stipulated by Brinckmann et al. (1988) that fatigue fractures of lumbar vertebrae would occur in vivo. However, such issues as time interval for load cycles, periods of rest, strain rate, maximum load in consecutive load cycles, etc., need to be addressed before these results can be applied. Further in vivo effects of tissue repair are not clearly understood.

This is an important finding and shows that the compressive strength is significantly lower under repeated loading of spine. This conclusion is also indirectly supported by Chaffin and Park (1973), who found increased risk of back pain in persons performing more that 150 lifts per day. Secondly, since strength of vertebral bodies decreases substantially with age, very low cyclic loads might induce fatigue fractures in the spine of older workers. However, from in vitro experiments, no direct conclusions can be drawn whether or not clinical symptoms originate in vivo from end plate fractures. This is a topic which needs to be further investigated.

Though several researchers have developed various biomechanical models to study compressive and shear forces on low back from manual materials handling tasks, only a few studies have been reported on the relationship between the estimated compressive force and the observed incidence rates of low-back pain. Chaffin and Park (1973) reported that incidence rates for

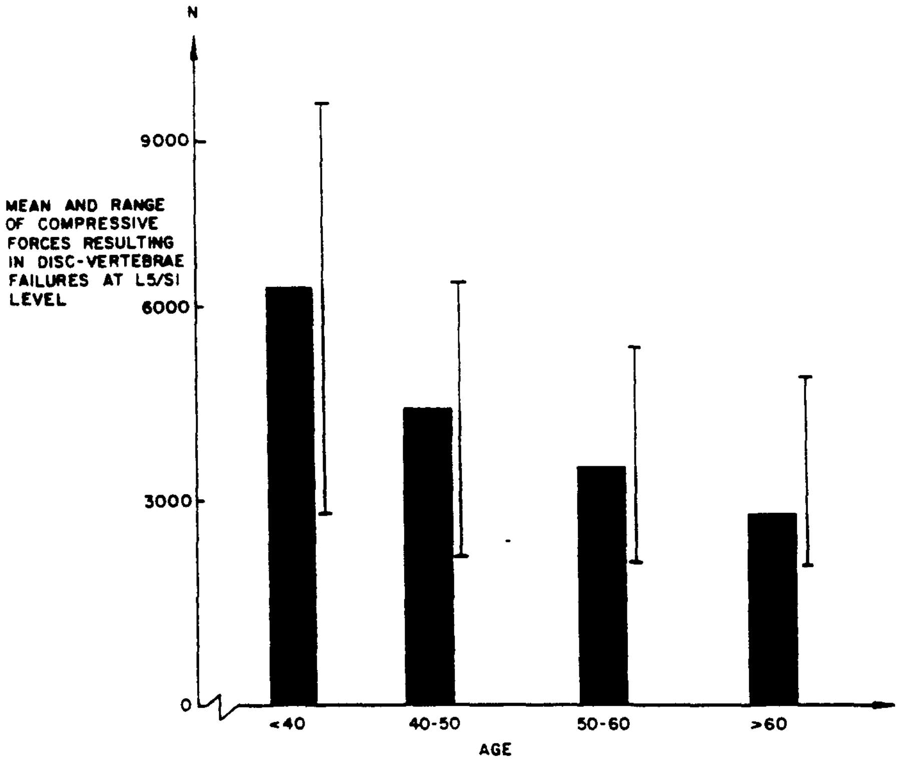


Figure 3.5: Composite of compression failure values obtained from cadavers of different ages. (Evans, 1959; and Sonoda, 1962)

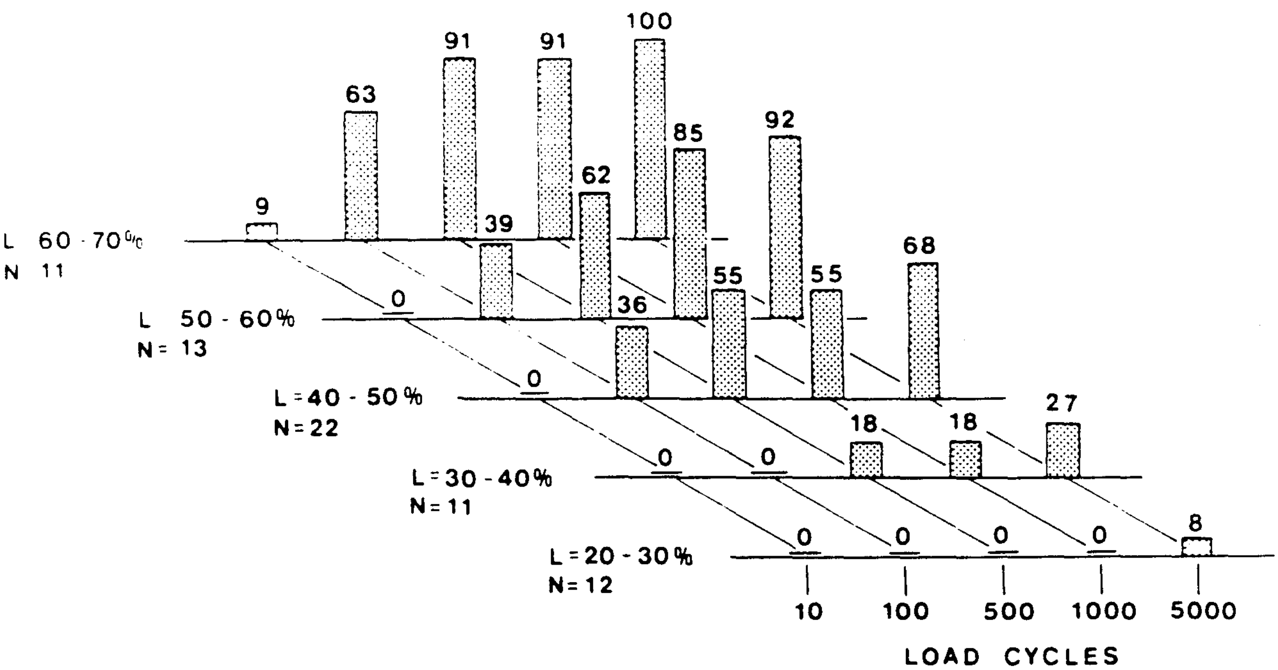


Figure 3.6: Probability of a motion segment to be fractured in dependence on the load range and the number of load cycles. (Brinckmann et al., 1988)

low-back pain were related to compressive forces on the L_5/S_1 disc (Figure 3.7). Further, the low-back pain incidence rates were significantly higher when the predicted compressive forces were greater than 6.37 kN (650 kg) (Figure 3.7). Bringham and Garg (1983) analyzed 109 cases of overexertion injuries over a period of three years in one company. There were ten cases of back injuries (including both muscular strains and disc injuries). In eight out of ten cases where muscular strains occurred, the average estimated compressive force was 5.34 kN. In two cases where disc injuries occurred, the average compressive force was 7.97 kN. Anderson (1983) reported a 40% increase in incidence rates for jobs requiring male compressive forces above 3.4 kN vs. below that level. Based on a study of 6912 incumbent workers in 55 industrial jobs in five industries, Herrin et al. (1986) concluded that "the biomechanical criterion of maximal back compression appears to be a good predictor not only of risk of back incidents but of overexertion injuries in general."

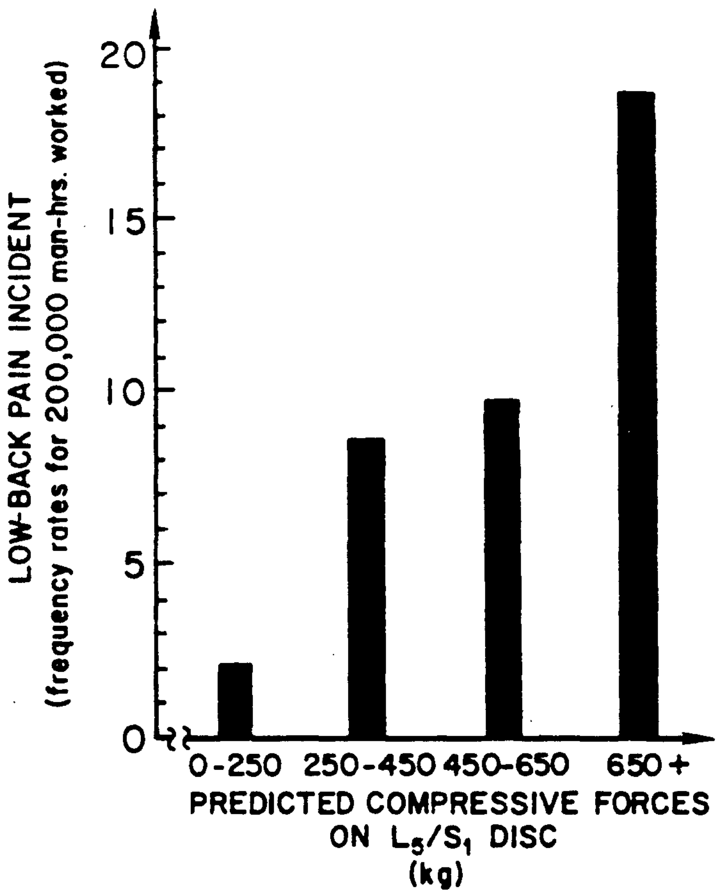


Figure 3.7: Relation between LBP and compressive Force. (Chaffin and Park, 1973)

Once again, from the above discussion it appears that it is difficult to provide a single value of spinal compressive strength that would be valid for all workers and for different work practices. However, it is apparent from Figures 3.4, 3.5 and 3.7 that **jobs which place more than about 6.37 kN (650 kg) of compressive force on the low back are hazardous to all but the healthiest of workers.** Only three out of twenty-five studies (or 12%) in Figure 3.4 reported mean fracture load greater than 6.4 kN. Similarly, an analysis of Brinckmann et al. (1988) data showed that only six out of thirty-five (or 17%) cadaver spinal columns had ultimate compressive strength of more than 6.4 kN under static load. Therefore, in terms of a specification for design, a much lower level should be recommended. It is clear from Figure 3.4 that only 12% of the studies have reported a mean compressive strength of less than 3.43 kN (350 kg). Brinckmann's et al. (1988) data showed that about 20% of the cadaver spinal columns had an ultimate compressive strength of less than 3.4 kN. **Thus, a value of 3.4 kN may not be protective for certain individuals, especially for those over 60 years of age (Figure 3.5) or for those suffering from degenerative changes.** Also, breaking strength under a given compressive load may be lower when the spine is subjected to additional shearing and rotational forces (Troup and Edwards, 1985). However, it is difficult to support a lower limit for ultimate compressive strength at this time, based on current knowledge of spinal compressive strength, recognizing in vitro limits may not be exactly applicable to in vivo, and given that several low-back injury mechanisms have been proposed. For repetitive lifting, those values may be lowered in the future as more knowledge is gained regarding fatigue fracture of the spine.

BIOMECHANICAL MODELS

From a biomechanics point of view, "safe" weight of a load depends upon: (i) capacity of muscles to develop appropriate tension (muscle strength), (ii) postural stability of the body (body balance), and (iii) compressive and shear forces on the spine assuming that stresses on ligaments and other musculoskeletal structures are not excessive. For example, the maximal weight for symmetric sagittal plane manual lifts in the erect standing posture usually depends upon the strength of the elbow and shoulder flexors even though the compressive forces on the spine may be small and one may have strong back and lower extremity muscles. Similarly, when lifting loads near the floor level or when lifting large objects away from the body, one may have sufficient muscle strength, but compressive forces on the lumbar spine are generally fairly high. When working in constrained work spaces or when the spine is laterally flexed, axially rotated or flexed (to reach for objects away from the body or grasp large objects), postural stability becomes a major problem in addition to compressive and shear forces and muscle strength requirements. Postural stability is also a major concern in those work situations where the foot traction is not sufficient.

Although intradiscal pressures (in the nucleus of the intervertebral disc) have been measured by Nachemson and Morris (1964), Nachemson (1965), Nachemson and Elfstom (1970), Andersson et al. (1977), Nachemson (1981) and Schultz et al. (1982) in laboratory studies under controlled conditions, such measurements are not recommended due to safety requirements and probably are not feasible in field studies. Most often, one has to simulate a given manual handling task using a biomechanical model to determine all three requirements of the task and, in particular, compressive force on the lumbar spine.

Several two and three-dimensional static and dynamic biomechanical models have been developed to determine stresses from manual handling tasks. Some of the static models include Chaffin (1969), Martin and Chaffin (1972), Garg and Chaffin (1975) and Anderson et al. (1985) and dynamic models include Grieve (1974), Troup (1977), Ayoub and El-Bassoussi (1978), Ekholm et al. (1982), Garg et al. (1982), Leskinen et al. (1983), Bejjani et al. (1984), Freivalds et al. (1984), McGill and Norman (1985) and Hall (1985), etc. However, most of these models used a single muscle equivalent to account for internal trunk muscle forces and resulting compressive and shear forces. The two-dimensional models appear to be satisfactory in analyzing two-handed symmetric sagittal plane exertions. Recent model validations of Garg and Chaffin's (1975) model showed a high correlation between the measured and the predicted strength--r^2 between 0.85 and 0.88 (Chaffin et al., 1987). However, for two-handed asymmetric exertion results were not as good--r^2 between 0.54 and 0.74 (Chaffin et al., 1987). The contemporary three-dimensional models of the trunk also showed that when the external moment is purely flexion (such as in symmetrical sagittal plane exertions), only the erector spinae muscles were active (Bean et al., 1988; Schultz et al., 1982). On the other hand, tasks involving extension, lateral bending and twisting moments tend to recruit many of the lumbar trunk muscles. Also, erector spinae muscles can be represented more easily by a single muscle equivalent than latissimus dorsi or the oblique abdominal muscles (Schultz et al., 1982).

More contemporary models of back include Schultz and Andersson (1981), Gracovetsky et al. (1981), Schultz et al. (1982), McGill and Norman (1986), Jäger (1987), Bean et al. (1988), Chen and Ayoub (1988), etc. These are three-dimensional biomechanical models of the lumbar back based on several muscle groups to more accurately reflect the muscle activities and compression and shear loads on the spine. Examples of typical simple coplanar and three-dimensional models of the trunk are shown in Figures 3.8 and 3.9. As stated earlier, more unknown internal forces and fewer equations of equilibrium make the three-dimensional models of the trunk statically indeterminate. Major assumptions are made in the estimation of internal muscle forces and these assumptions need to be validated against experimental measurements (such as myoelectric activity, intradiscal pressure, etc.). Among other assumptions, these models include a biomechanical objective function such as minimization of compressive force, shear force, contraction intensity in any muscle, higher muscle contraction intensity, etc., or a combination of these variables. At present, individual muscle models are of limited practical value due to mechanically indeterminate systems and because the precise relation between the mechanical and electrical output of a muscle is uncertain (Ortengran and Andersson, 1977).

The model by Schultz and Andersson (1981) has been validated against myoelectric activity, intradiscal pressure and intra-abdominal pressure. Validation studies by Schultz and Andersson (1981) and Schultz et al. (1982a, b) for light to moderate exertions reported a correlation coefficient of 0.91 to 0.98 between mean intradiscal pressure and mean spinal compression. For less strenuous tasks, these studies and Schultz et al. (1987) reported correlation coefficients between myoelectric activities and muscle contraction forces of 0.6 to 0.99 for erector spinae, 0.70 to 0.97 for rectus abdominis and 0.34 to 0.93 for abdominal oblique muscles. The correlation coefficient between the measured intra-abdominal and the measured disc pressure was rather low (0.36; Schultz et al., 1982). For more strenuous tasks, Schultz et al. (1987) concluded that the model predictions were inadequate. The study concluded that during strenuous exertions: (i) substantial antagonistic muscle contractions sometimes occurred, (ii) intra-abdominal pressurization might sometimes have contributed substantially to the maintenance of structural equilibrium, and (iii) the ligamentous tissues of the trunk seemed sometimes to develop substantial passive resistances to bending and twisting moments.

Most biomechanical models assume negligible antagonistic activity either directly or indirectly through choice of an objective function. The model by Bean et al. (1988) accounted for antagonistic muscle activity by using a double linear programming approach. However, their model needs to be validated against indirectly measured internal forces (intradiscal pressure and myoelectric activity) for different levels of physical exertions and body postures. Bean et al. (1988) suggests that the antagonistic muscle activity cannot be assumed to be zero if more than one external moment is present since the antagonistic activity may be agonist in resisting some other moment. Electromyographic studies have also confirmed that there is a significant antagonistic muscle activity present in certain exertions (Portnoy and Morin, 1956; Morris et al., 1962; Ortengren and Andersson, 1977; Pope et al., 1987; Zetterberg et al., 1987; Schultz et al., 1987).

Farfan (1973), Gracovetsky et al. (1981), Gracovetsky et al. (1985), Gracovetsky and Farfan (1986) suggest that the posterior ligamentous system plays an important role in reducing

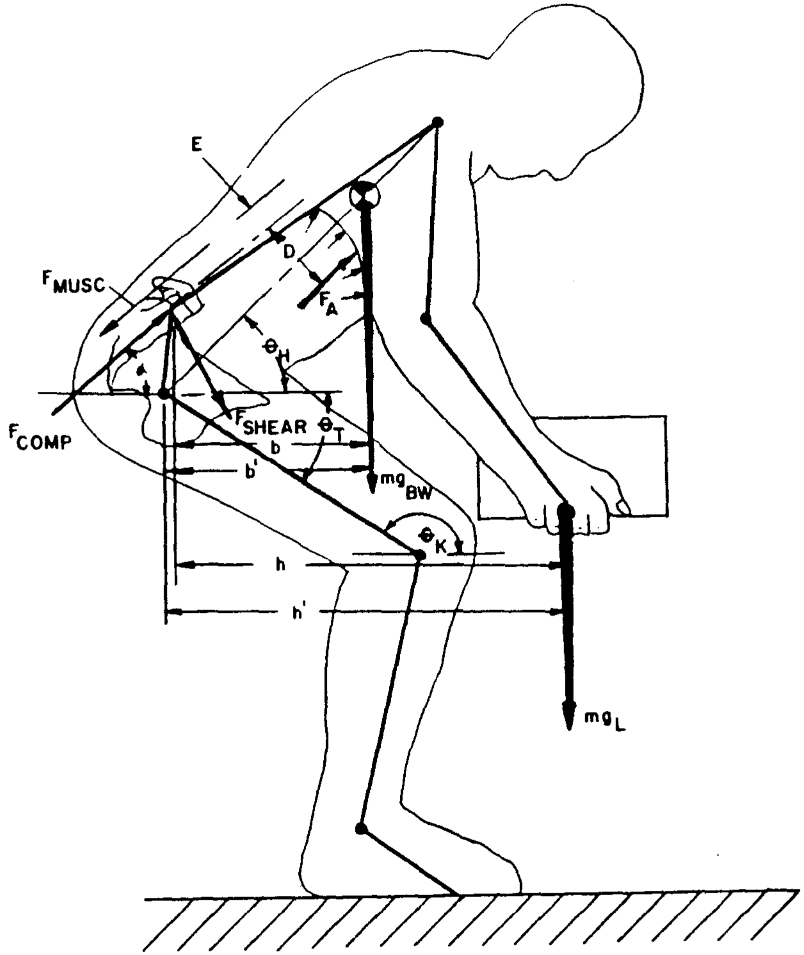


Figure 3.8: Simple low-back model of lifting as adapted by Chaffin (1975) for static coplanar lifting analyses.

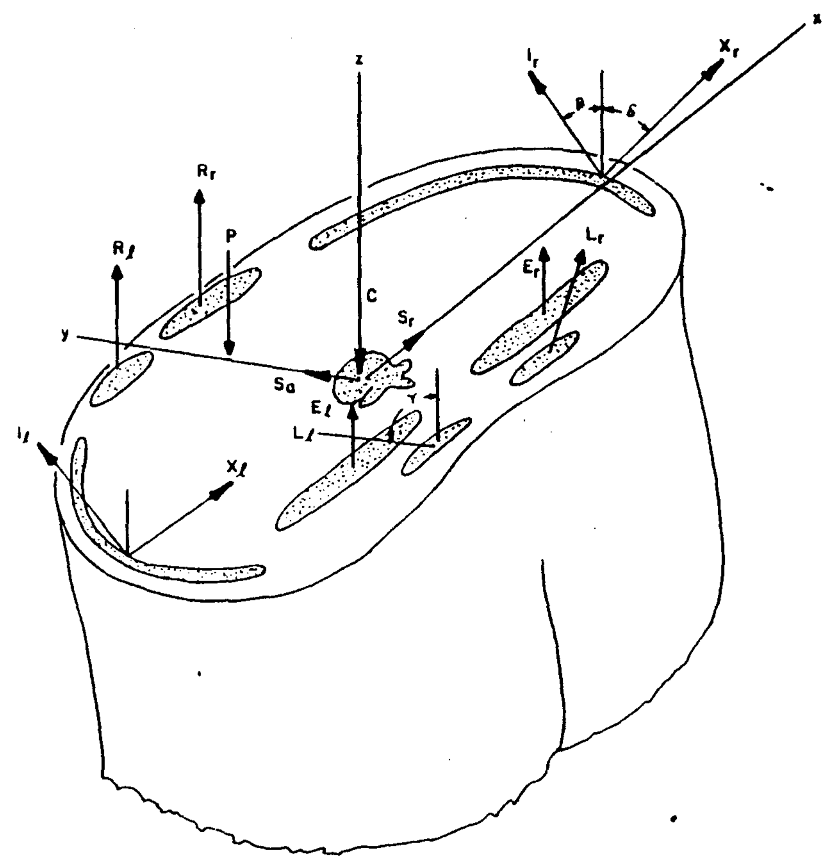


Figure 3.9: Schematic diagram of the model trunk cross section at the third lumbar level. The biomechanical analysis of the internal forces requires that the six components of the net reaction and the intra-abdominal pressure force (P) be known from experimental measurements. The ten unknown muscle forces are computed to provide the three net reaction moment components and the minimum spine compression force. Knowledge of the two remaining net force components is then used to compute the two motion-segment shear forces. (Schultz et al., 1982)

compression at the intervertebral joint because of the angled attachment and larger moment arm. Theses authors suggest that it is important to take advantage of the capability of ligaments to partially relieve the trunk musculature of their role in counteracting the moment on intervertebral joint, especially when the lift is performed at speed with flexed back (slightly rounded spine). Using a mathematical model, Gracovetsky et al. (1981) showed that a given task could be performed by utilizing any number of combinations of muscle power and ligament tension and when support of the external load was shifted from muscle to ligament, there was a decrease in compression load at the disc (Gracovetsky et al., 1985; Gracovetsky and Farfan, 1986). However, Tesh et al. (1977) showed that the stabilization action of thoracolumbar fascia was relatively small in a fully flexed position. McGill and Norman (1986) found that ligaments played a very minor role in the lifts studied and their study could not confirm the hypothesis advanced by Gracovetsky et al. (1981, 1985). These authors concluded that posterior ligaments are not a significant component of the lifting spinal mechanism during stressful, sagittal plane lifting. Anderson et al. (1985) found that for sagittal plane two-handed lifting, the restorative moment due to ligament resistance was fairly small as compared with those due to muscle contraction and abdominal pressure. Moreover, under a 500 N load the ligaments played a much smaller role than under the no-load condition. The study concluded that typical lifting tasks could lead to excessive disc compressive forces, muscle moment generation requirements, and possibly lumbodorsal fascia strains. Adams and Hutton (1980), based on a cadaver study, concluded that the posterior ligaments (supraspinous, interspinous, capsular ligaments and ligament flavum) could resist less than half of the weight of the trunk in full flexion. The rest of the moment would be balanced by the passive resistance offered by the lumbodorsal fascia and extensor muscles of the spine. In this regard, though it has been shown that erector spinae may be electrically silent in full flexion (Floyd and Silver, 1955), they could still provide tension passively.

In summary, detailed three-dimensional biomechanical models of the trunk, as compared to simple models, provide more accurate representation of trunk musculature and more accurate prediction of muscle contraction and compressive and shear forces on the lumbar spine, especially for tasks involving axial rotation and/or lateral flexion. However, under what conditions antagonistic muscle activity and ligamentous tissue play an important role and what is the extent of their contribution to restorative moments and spinal compressive forces are not fully understood. In addition to these two, there are other issues which should be considered in interpreting predictions from biomechanical models. Generally, the model validations have been performed in a laboratory under carefully controlled body postures, such as strapping the pelvis to a support board to stabilize the lower body segments (Schultz et al., 1982a, b;

Ortengren et al., 1981). When applying these models to manual tasks in industry, it is probably not possible to control the body posture precisely and it may be difficult to obtain precise three-dimensional postural data. Also, subjects instinctively change the direction of the applied force from true vertical or horizontal in order to maximize the actual lifting, pushing, or pulling force (Garg et al., 1983; Grieve and Pheasant, 1981). Ortengren et al. (1981) showed that such variables as lifting method, height, and symmetric vs. asymmetric loading body postures, etc., might affect the relationship between intradiscal pressure, myoelectric activity and intra-abdominal pressure. Moreover, muscle recruitment patterns may vary from person to person and from time to time in any one person (Donisch and Basmajian, Ortengren et al., 1977; Schultz et al., 1982). Therefore, detailed biomechanical model validations are needed. In spite of these reservations, for the present it is **believed that the biomechanical models are very useful in determining the etiology of low-back pain and in quantifying stresses on various parts of the body and, in particular, on the lumbar spine. This is an important topic which needs to be further investigated.**

EFFECT OF DYNAMIC FACTORS ON SPINAL STRESSES

One criticism of static biomechanical models is that lifting is a dynamic activity. Static models tend to underestimate forces and moments because the inertial loads imposed by dynamic actions are ignored. For example, Garg et al. (1982) reported that peak compressive forces at the L_5/S_1 disc, peak dynamic moments and forces in the back muscles for lifting maximum acceptable weights (about 30 kg) were two to three times greater than those based on the static biomechanical simulation. Also, the study reported that the peak compressive forces occurred at about 0.225 to 0.628 S from the beginning of lift (Figure 3.10). Leskinen et al. (1983) reported that when lifting a 15 kg box from 10 cm off the floor the predicted forces on the lumbar spine were 33 to 60% higher with the dynamic model depending on which lifting technique was used (Figure 3.11). McGill and Norman (1985) reported 19% greater moment with the dynamic model as compared to the static model for lifting an 18 kg load. Freivalds et al. (1984) concluded that dynamic effects increased the effect of static load by as much as 40%. These and other studies (Troup et al., 1983; Hall, 1985; Bush-Joseph et al., 1988; etc.) have shown that dynamic factors significantly increase the stresses on the spine when lifting a load. Definitely, the difference between the static and the dynamic analysis will depend upon the acceleration of the load being lifted, the upper limbs and the trunk. Differences are smaller with slower lifts.

Clearly, inclusion of dynamic factors in biomechanical models is necessary to more accurately estimate the moments and forces on various body joints, including the low back. However, at

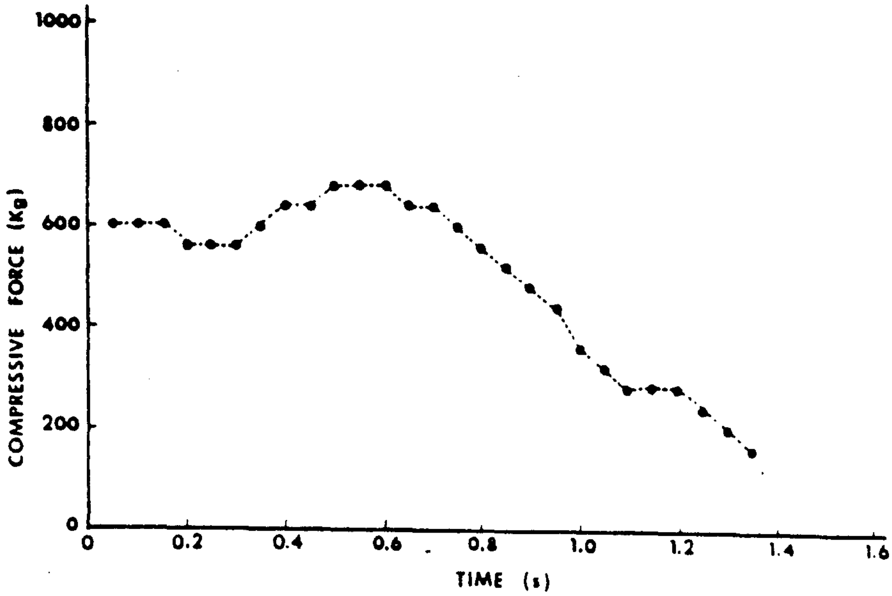


Figure 3.10: Estimated compressive forces at the L_5/S_1 disc from a dynamic biomechanical simulation of lifting a 38 cm box with handles for a typical subject versus time into lift. (Garg et al., 1982)

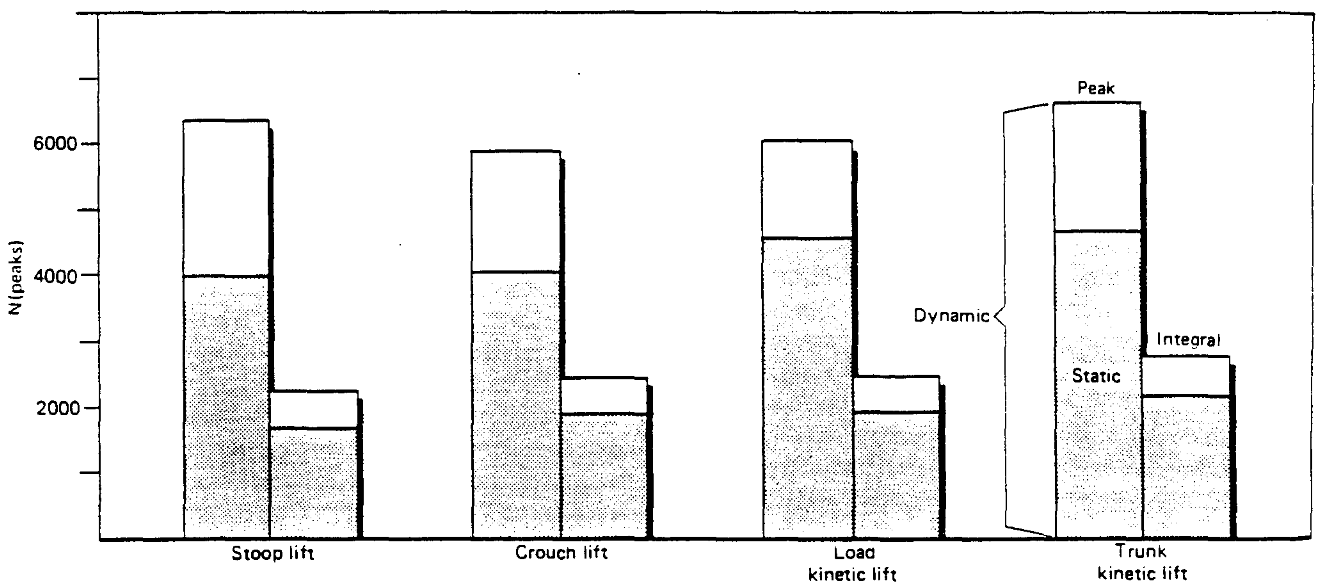


Figure 3.11: Mean peak lumbosacral compressions (N) and mean compression-time integrals (Ns) for 20 males lifting a 15-kg box, comparing calculations based on static and dynamic models, the latter taking inertial factors into account. (Leskinen et al., 1983b)

present dynamic biomechanical models are of limited practical use in analyzing stresses from manual handling jobs and in designing workplaces. Ultimate compressive strengths of lumbar vertebral bodies, such as those presented in Figures 3.4 and 3.5, are not available under different dynamic loading conditions. Volitional muscle strengths (muscle moment capacities) and tissue limits for various body joints have not been systematically studied as a function of angular velocity and acceleration. Thus, at present it is not possible to compare the task produced moments and forces with the allowable muscle moment capacities and compressive forces. To some extent, predictions of compressive forces on the lumbar spine, even under static loading conditions, depend on the assumptions made in determining internal forces in muscles and ligaments, etc. Inclusion of dynamic factors may compound these errors. Also, it is difficult to obtain postural data in three dimensions as a function of time, especially in an industrial setting. These issues need to be better understood before workplace guidelines can be proposed based on a dynamic analysis. It is concluded that for the present one has to rely on static biomechanical analyses to determine manual materials handling guidelines.

INTRA-ABDOMINAL PRESSURE

It is common at the start of a manual handling activity to hold the breath and close the glottis. The muscles of the abdominal wall and pelvic floor contract and pressure increases in both thoracic and abdominal cavities. Increased pressure in the thorax stiffens the rib cage, provides a stable base for the activity of the upper limbs and supplements the thoracic erector spinae musculature (Davis and Troup, 1964; 1966). It is believed that a pressure increase in the abdominal cavity (intra-abdominal pressure) appears to supplement the lumbar extensor mechanism, particularly when the spine is flexed (Figures 3.8 and 3.12). An increase in intra-abdominal pressure creates a force tending to extend the spine.

The possible effects of intra-abdominal pressure in reducing the compressive force are not clear due to current controversy surrounding this mechanism (Gracovetsky et al., 1981; Schultz et al., 1982; Bogduk and McIntosh, 1984; Pope et al., 1984; Marras et al., 1984). Gracovetsky et al. (1985) believe that the primary purpose of abdominal pressurization is to ensure the proper shape (for maximum efficiency of lumbodorsal fascia) and to support the hoop tension generated by the contraction of internal abdominis and transverse abdominis muscles. The rectus abdominis counterbalances the action of intra-abdominal pressure on the diaphragm and the pelvic floor. They conclude that, "... it is apparent that any direct relief of the pressure of the disc by intra-abdominal pressure per se is negligible." Pope et al. (1984) state that reduction of muscle extension moment and compressive force by intra-abdominal pressure is controversial

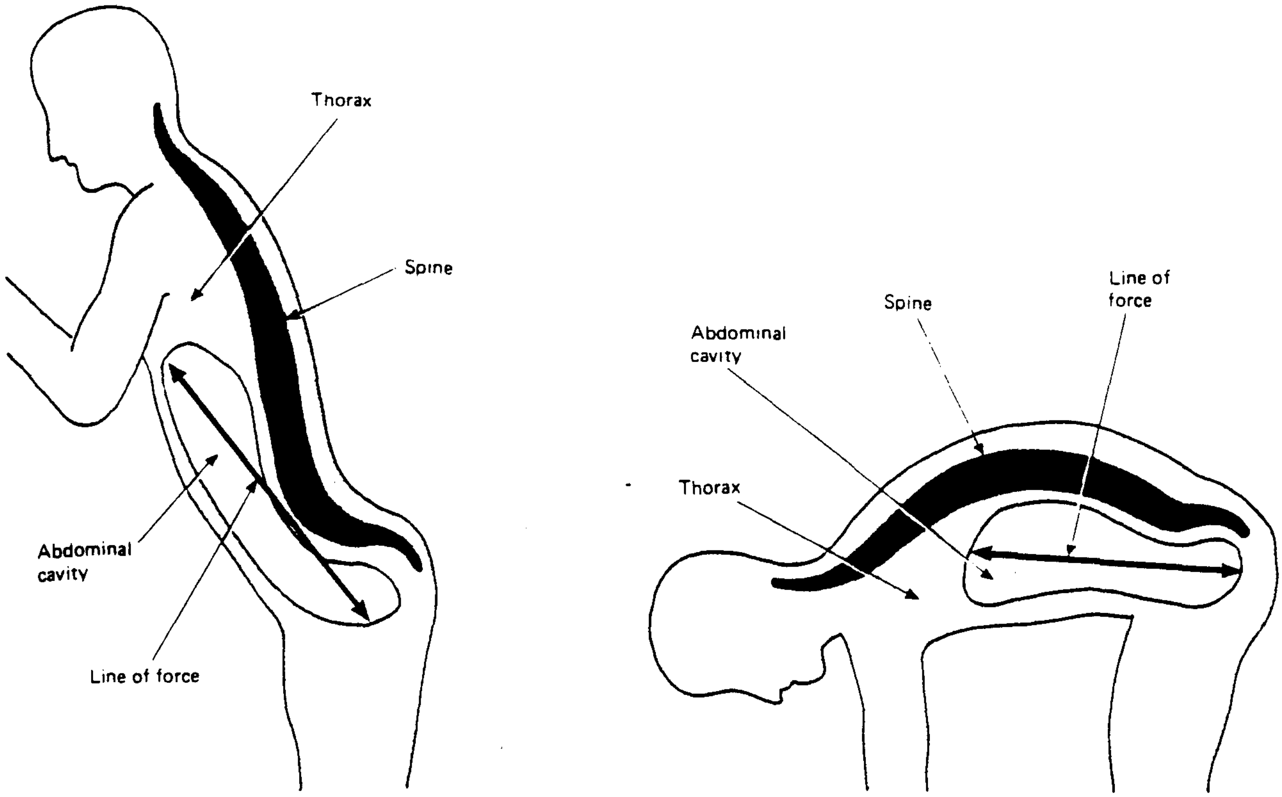


Figure 3.12: Lines of force of IAP in flexed and extended positions. (Troup, 1965)

since the abdominal muscles must contract to produce the intra-abdominal pressure and this would produce a flexion moment. McGill and Norman (1986) found that the extensor force and moment created from intra-abdominal pressure did not offset the compression force generated by necessary abdominal activation. Troup (1979) believes that the oblique muscles are more axially oriented and may to some extent oppose the effect of intra-abdominal pressure increase. Gilberton et al. (1983) showed that an increase in intra-abdominal pressure might not decrease the activity of the dorsal musculature in flexed or axially rotated position. As stated in Chaffin and Andersson (1984), asymmetric loading of the trunk can create large stresses on the lumbar spine without simultaneously increasing intra-abdominal pressure.

Marras et al. (1984) found substantial activity in the internal oblique, external oblique and rectus abdominis muscles under both static and isokinetic conditions but no relationship between the intra-abdominal pressure and the abdominal musculature. Moreover, they reported that in many high velocity isokinetic tests, there was an onset of delay in the production of intra-abdominal pressure and muscle moment, and this lag increased with increasing velocity. In many high velocity lifting tasks, the intra-abdominal pressure terminated before the moment production began. Marras et al. (1984) concluded that intra-abdominal pressure might not possess any load relieving capabilities. Legg (1981) found that the intra-abdominal pressures developed in the submaximal lifting were not influenced by fatigue or training of the abdominal muscles. Thus, it appears that the training and fatigue of abdominal muscles do not play an important role in the risk of back injury.

Intra-abdominal pressure response to stoop lifting can be divided into an initial peak (the snatch pressure), a lower sustained pressure while the load is raised to the target level (the lift pressure), and a further peak associated with placement of load (Davis, 1959; 1981) (Figure 3.13). The peaks are believed to be associated with extra moment required to accelerate the load and the trunk at the beginning and end of lifting, and coincide with the accelerative phase (Davis and Troup, 1964). Intra-abdominal pressures increase during pushing and pulling also. Pressure increase is in proportion to the speed of lifting and the weight of the load (Davis, 1959; Davis and Troup, 1959), and, thus, is in proportion to the stress imposed on the spine (Andersson et al., 1979). Davis et al. (1964) observed that jerk movements produced higher intra-abdominal pressure than slow, smooth lifting motions. As an extensor mechanism, intra-abdominal pressure is most effective during the short periods of maximal handling effort, especially when the spine is flexed. When the spine is extended (as in lifting above the shoulder), intra-abdominal pressure is less effective due to a reduction in the length of lever arm (Davis and Troup, 1965; Figure 3.12).

Intra-abdominal pressure has been found to correlate satisfactorily with the calculated compressive forces and the moments on the lumbar spine in the sagittal plane (Davis et al., 1965; Marras et al., 1984). Andersson et al. (1977) and Ortengren et al. (1981) showed that intra-abdominal pressure was linearly related to both intradiscal pressure and the myoelectric activity of erector spinae during static exertions. However, Schultz et al. (1982) reported a correlation of 0.36 between the measured intra-abdominal pressure and measured intradiscal pressure and 0.24 between the intra-abdominal pressure and the predicted compressive force. Davis et al. (1981) stated that lifting with the spine rotated resulted in higher levels of intra-abdominal pressure. Andersson et al. (1977) reported that the intra-abdominal pressure increased when the trunk was loaded in lateral flexion as well as in rotation. Davis et al. (1983) reported higher intra-abdominal pressure with stoop lift than with crouch lift, but Troup et al. (1983) found the opposite to be true. For a given task intra-abdominal

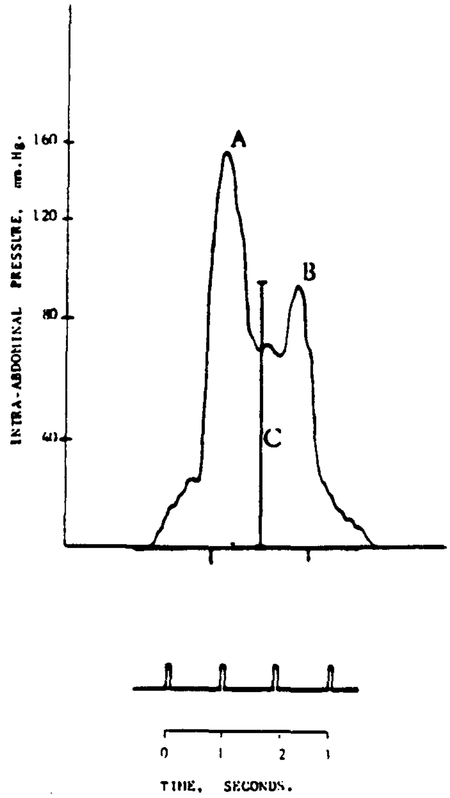
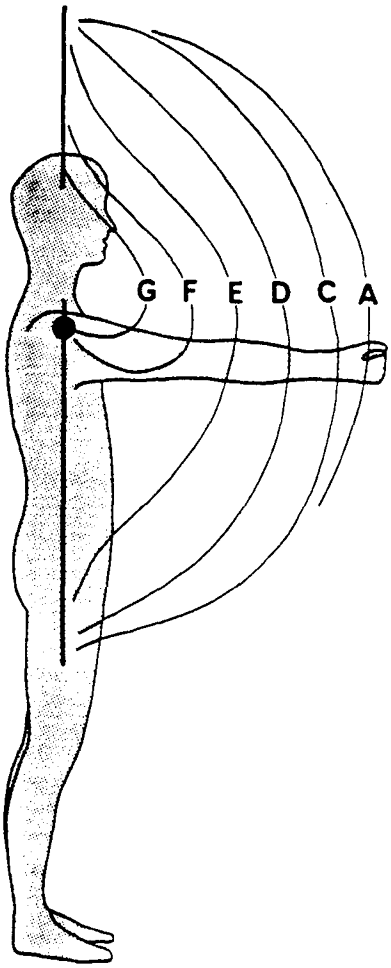


Figure 3.13: Intra-abdominal pressure record of subject lifting a load of 25 kg from ground level to 1.11 m using the straight full stoop technique. A) First peak pressure = 160 mm Hg; B) second peak pressure = 94 mm Hg; C) mean pressure = 95 mm Hg; D) pressure quotient (mean pressure x time of lift) = 170 mm Hg.sec. (Davis, 1981)

pressure is higher in older people--above 50 years of age (Davis, 1981; Davis and Stubbs, 1980)--and in those with back pain (Fairbank et al., 1980).

An association between high levels of intra-abdominal pressures (100 mm of Hg and above) and a high prevalence of back injuries at work has been established (Davis and Stubbs, 1978; Davis and Shephard, 1980; Nicholson et al., 1981). Davis and Stubbs (1980) published contour maps for acceptable levels of force without exceeding an intra-abdominal pressure of 90 mm of Hg (Figure 3.14). Coefficients of variation were also calculated from 1080 individual observations to determine the reliability of the method. The mean coefficient of variation for all activities was 32.3% (Davis, 1981) as compared to 50-100% reported by Ortengren et al. (1977). Davis (1981) concluded that intra-abdominal pressure measurements are useful determinants of spinal stress.



Force levels (kg) for different age groups

	AGE GROUPS		
	≤ 40 MALE	41–50 MALE	51–60 MALE
A	10	10	9
B	11	11	10
C	12	12	11
D	15	15	13
E	20	20	18
F	25	25	22
G	30	30	27

ONE-HANDED UPWARD, VERTICAL FORCES (INCLUDING LIFTING) when standing or squatting with the trunk reasonably upright.

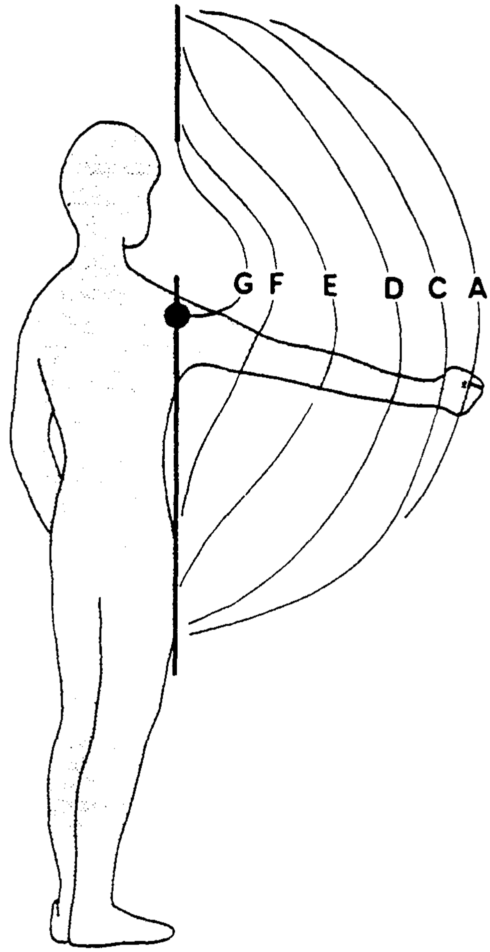


Figure 3.14: Contour maps showing the distances from the body at which acceptable levels of force can be applied in a variety of postures with the trunk erect, without exceeding IAP of 90 mm Hg. (Davis and Stubbs, 1980)

Morris et al. (1961), using a mathematical model of forces on spine, estimated that the increased intra-abdominal pressure reduced the load on the spine by 30%. Davis and Troup (1965) concluded that the intra-abdominal pressure could reduce lumbosacral compression by as much as 25-30% and Eie (1966) by as much as 40%. Subsequent studies have estimated a lower load reduction level. Thomson (1988) estimated that the intra-abdominal pressure is capable of reducing the compressive load on the lumbar spine by about 20% during heavy lifting. Chaffin (1982) concluded that when assumed to be a well developed reflex in an individual, it appeared that the intra-abdominal pressure could relieve about 15%-20% of the lumbar compression during normal weight lifting actions. Schultz et al. (1982) reported that the mean relief of compression on the spine due to intra-abdominal pressure was 14.5% (range = 4 to 35%), and this value did not show any consistent relationship with the load imposed on the spine by the task performed. The authors also concluded that the intra-abdominal pressures were not large and seldom had a major influence on the overall mechanics of the trunk. However, in a subsequent study, Schultz et al. (1987) discussed the possibility that the intra-abdominal pressurization might sometimes contribute substantially to the maintenance of structural equilibrium when performing strenuous tasks. Using a biomechanical model, Garg and Herrin (1978) reported that the intra-abdominal pressure could reduce the compressive load on the lumbar spine from 8.7% to 20.3% for stoop lift and from 2.6% to 11.3% for squat lift for lifting loads ranging from 0 to 70 kg. Recently, Anderson et al. (1985) concluded that abdominal pressure could never relieve more than about 10% of the compressive force.

In summary, at present the role of intra-abdominal pressure in relieving compression on the spine appears to be controversial. It is not quite clear under what circumstances intra-abdominal pressure plays an important role and what is the magnitude of compression relief provided by the intra-abdominal pressure. It appears that the intra-abdominal pressure may reduce about 10%-20% compressive load on the lumbar spine during strenuous exertions.

ASYMMETRIC LIFTING

Asymmetric lifting occurs when the load is lifted off the mid-sagittal plane with one or both hands, when the spine is axially rotated (twisted), or laterally flexed, or a combination of these. More often the spine is subjected to forward flexion accompanied by axial rotation and lateral flexion. It is generally believed that asymmetric lifting in more hazardous to the musculoskeletal system than symmetric lifting because of the combined effects of flexion, axial rotation and lateral bending. Unfortunately, there are only a few studies of asymmetric load handling because of experimental and biomechanical complexities associated with three-dimensional force analysis. At the same time twisting of the spine while lifting is common in workplaces where origin and destination are oriented at an angle to one another, where there is inadequate room to use a step turn, where lifting is done across the body as in swinging bags or boxes up from a low level, or where work is done in obstructed workplaces or on littered floors to maintain body balance (Rodgers, 1985). Sometimes workers prefer to twist and/or laterally flex their spines because it is faster than lifting symmetrically in the sagittal plane with turns and one or more steps involved (Garg, 1986).

Farfan (1973) showed that spinal mechanism failed either by excessive compressive load or excessive torsional load. Farfan (1985) and Kirkaldy and Farfan (1981) estimated that approximately 65% of low-back cases are from torsional injury and 35% from compression injury (based on citations in Gracovetsky, 1986). As stated in Fiorini and McCammond (1976), Ghormely (1951) arbitrarily related low-back pain to degenerative disc disease in about 22% of cases and due to degenerative changes in the synovial joints in about 26% of cases. Facet joint pain has been estimated to be a common cause for chronic low-back pain (Pope et al., 1984). Frymoyer et al. (1980) reported that patients who reported low-back pain had occupations that required more repetitive lifting, pulling and twisting.

The torsional injury is basically a rupture of ligamentous tissue, the annulus, and the associated damage to the facet (Gracovetsky, 1986). The study also states that torsional injury is by far the most damaging and the ligamentous injuries do not heal quickly. In a torsion injury, the joint never regains full strength and therefore the same joint is most likely to fail with repeated torsion (Farfan, 1985). Torsional injuries involve peripheral damage to both the disc and the facet joints occurring simultaneously (Farfan, 1985). The annulus is avulsed from the endplate and its laminae become separated while the nucleus and endplate remain intact. At a later stage, the annulus develops radial fissures and the changes in facet joints are severe and the intervertebral joint may become unstable (Farfan et al., 1970; Farfan, 1985; Gracovetsky, 1986). The torsional deformity can cause the neural elements in the intervertebral canal to displace to one side and stretch the nerve roots (Farfan, 1985). Symptoms may arise form intrusion of disc (annulus) into the canal, stretched nerve roots, constriction of neural elements by torsional deformation (stenosis) and/or instability (Farfan, 1985).

Mathematical models demonstrate that torsion produces a stress concentration at the region of posterior lateral annulus which is a common site of disc degeneration (Hickey and Hukins, 1980; Pope et al., 1984; Gracovetsky and Farfan, 1986). Based on a cadaver study, Farfan et al. (1970) estimated that about 90% of the torque strength of an intervertebral joint was provided by its disc (annulus) and the two joints between the articular process. The annulus fibrosus provided 40%-50% of the torsional resistance to twisting of the lumbar vertebrae. The nucleus being a gel, can provide little or no resistance to torsion. The authors proposed that the cohesion between the laminae gave the disc its torque strength. Therefore, damage to annulus as a result of disease, fracture or surgical assault must have serious consequences for the whole joint (Farfan et al., 1970). The authors speculated that with each successive injury, deeper and deeper annular layers might be affected, and gradually a communication between the nucleus and the outside might be established, forming the radial tears typical of natural disc degeneration. Further, with the loss of cohesion between the laminae, the articular facets will have to provide a greater portion of resistance to torsion. This may increase the risk of disc degeneration. Based on a study of 42 cadaver lumbar motion segments, Schultz et al. (1979) reported that the segments were least flexible in torsion (as compared to extension, lateral bending and flexion), destruction of posterior elements (facet joints and ligaments) resulted in significantly increased motions in torsion and extension (with little effect in flexion and lateral bending), intradiscal pressure increases were relatively small in torsion and extension (as compared to flexion and lateral bending) and significantly large intradiscal pressure increases occurred in torsion and extension when posterior elements were removed. In short, posterior elements of a motion segment played a significant role in torsion and extension and little or no role in flexion and lateral bending. Skipor et al. (1985) showed marked increases in rotation when posterior ligaments were severed leaving the facet joint capsule intact. Based on loading of cadaver lumbar intervertebral joints simultaneously in torsion and compression, Adams and Hutton (1981) showed that torsion of the lumbar spine was resisted primarily by the apophyseal joint that was in compression, although the intervertebral disc played a major role. They showed that the compression facet was the first structure to yield at the limit of torsion. The study suggested that a local collapse of the articular cartilage or subchondral bone of the articular facet might have occurred. Further, the intervertebral disc was subjected to relatively small stresses and strains (1/3 of its maximum angle) in the physiologic range of torsion because of the protection offered by the compression facet. The authors also suggested that torsion seemed unimportant in the etiology of disc degeneration and prolapse as rotation is produced by voluntary muscle activity. Based on biomechanical analysis, Fiorini and McCammond (1976) also supported Adams and Hutton's observations. They inferred that if a person twisted while lifting and bending, the pressure on one facet would increase considerably, while it decreased on the other. The interarticular facet joints, being small and joined together at the periphery by a very thin and delicate capsular ligament, are unstable and vulnerable to injury (Fiorini and McCammond, 1976). Since the synovial membrane of the facets is highly vascular and innervated with sensory nerves, the articular facets are likely to cause considerable pain (Fiorini and McCammond, 1976).

Farfan et al. (1970) reported that the average torque at failure for intact whole intervertebral joint was 88 Nm (range = 85 to 101 Nm) for those with the normal discs and 54 Nm (range = 33 to 68 Nm) for those with degenerated discs. The mean rotations at failure were 22° and 14° for normal and degenerated discs, respectively. However, the authors suggested that the stress-strain relationship was exponential and there was probably a change in behavior of the disc at 3 degrees of rotation. Further, the study concluded that the intervertebral joint might sustain injury at much smaller rotation of 2 to 5 degrees. This would correspond to an axial moment of 11 to 23 Nm (Farfan et al., 1970). This is consistent with the findings of Adams and Hutton (1981), who reported that most specimens reached the limit of torsion with a torque of 10 to 30 Nm and with a rotation of between 1° and 3°. However, Adams and Hutton (1981) found that the more degenerated joints had greater angles of rotation before failure (1.2° with grade 1 discs and 6.7° with grade 4 discs). Based on a mathematical model of the disc, Hickey and Hukins (1980) computed 16 Nm as the maximum torque that would not damage the annulus. This would correspond to a 3° torsion angle or 4% stretch in the annular fibers.

The intra-abdominal pressure, myoelectric activity of trunk muscles, intradiscal pressure and compressive force have also been studied for asymmetric activities. In general, high levels of antagonistic activity in both abdominal and posterior back muscles has been reported in lateral flexion and axial rotation (Andersson et al., 1977; Kumar, 1980; Ortengren et al., 1981; Schultz et al., 1982; Pope et al., 1987; Seroussi and Pope, 1987; Zetterberg et al., 1987; Schultz et al., 1987). These studies also found higher myoelectric activity on the side contralateral to the load in the lumbar region though there was significant muscle activity on the ipsilateral side. Seroussi and Pope (1987) reported high correlations between the difference of right and left erector spinae myoelectric activity and the frontal plane moment arm. Since several muscles are active under asymmetric lifting conditions, it is difficult to quantitatively compare the myoelectric activity for asymmetric exertions with those for symmetric exertions. However, Pope et al. (1987), in general, showed greater myoelectric activity at a given percentage of maximum voluntary contraction (MVC) when the trunk was rotated by 30° as compared to the neutral position. They also reported that the ability to develop maximum torque was enhanced if the subject was prerotated away from the direction in which the torque was being developed. Andersson et al. (1977) reported higher intra-abdominal pressure and myoelectric activity when the trunk was loaded in rotation than in lateral flexion.

Ortengren et al. (1981) showed higher intradiscal pressure for a given level of myoelectric activity for asymmetric loading positions than that for symmetric loading positions. Andersson et al. (1977) reported that intradiscal pressure and the intra-abdominal pressure both increased when the trunk was loaded in lateral flexion (20°) as well as in axial rotation (45°) (Figure 3.15). The measured pressures were higher when the trunk was loaded in rotation than in lateral flexion (Figure 3.15). However, it is not clear from Figure 3.15 how much increase in intradiscal pressure is due to trunk flexion and how much increase is due to axial rotation. Bean et al. (1988),

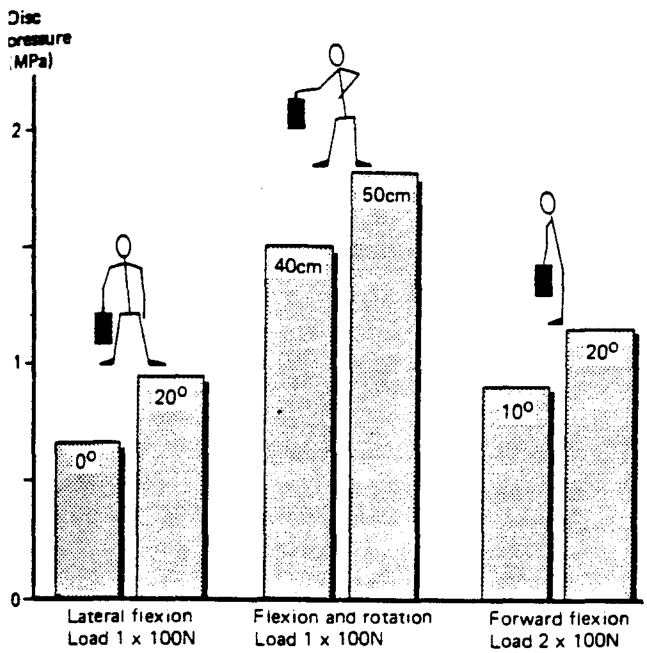


Figure 3.15: Intravital disc pressures (MPa) during static loading in left lateral flexion with a load of 100 N (10 kg) in the left hand; with combination of flexion and rotation and a load at a distance of 40 and 50 cm from the center of the body respectively; and, finally, in forward flexion with 100 N (10 kg) in each hand at 10 and 20° respectively. (Nachemson, 1980)

using a double linear programming approach, estimated compressive force for holding a 222 N weight on a circular arc of 38 cm from the L_5/S_1 disc center at 0, 30, 60, and 90° angular displacements. Estimated values from their graph showed that the compressive forces at 30, 60 and 90° were about 14%, 4% and 36% higher, respectively, than those at 0° (mid-sagittal plane). Figure 3.16 shows percentage increase in compressive force compared to 0° for lifting a load of 400 N at a vertical height of 81 cm for three different horizontal distances (Chaffin, 1989; personal communications). In each of these postures, the hand to hand separation distance was held at 51 cm and an erect torso was maintained. However, these findings are not supported by Schultz et al. (1982), who found that the tasks involving axial rotation or lateral flexion did not load the spine and trunk muscles significantly more than trunk flexion or holding of weights in front of the body. Neither the myoelectric measurements nor the model analysis (compressive, shear and trunk muscle contraction forces) could support the hypothesis that tasks involving axial rotation, lateral flexion, or a combination of these with forward flexion unduly load the lumbar structures. According to Schultz et al. (1982), it is the flexion moment (such as due to outstretched arms or trunk flexion) rather than the twist or lateral bending that is responsible for the large loads on the spine.

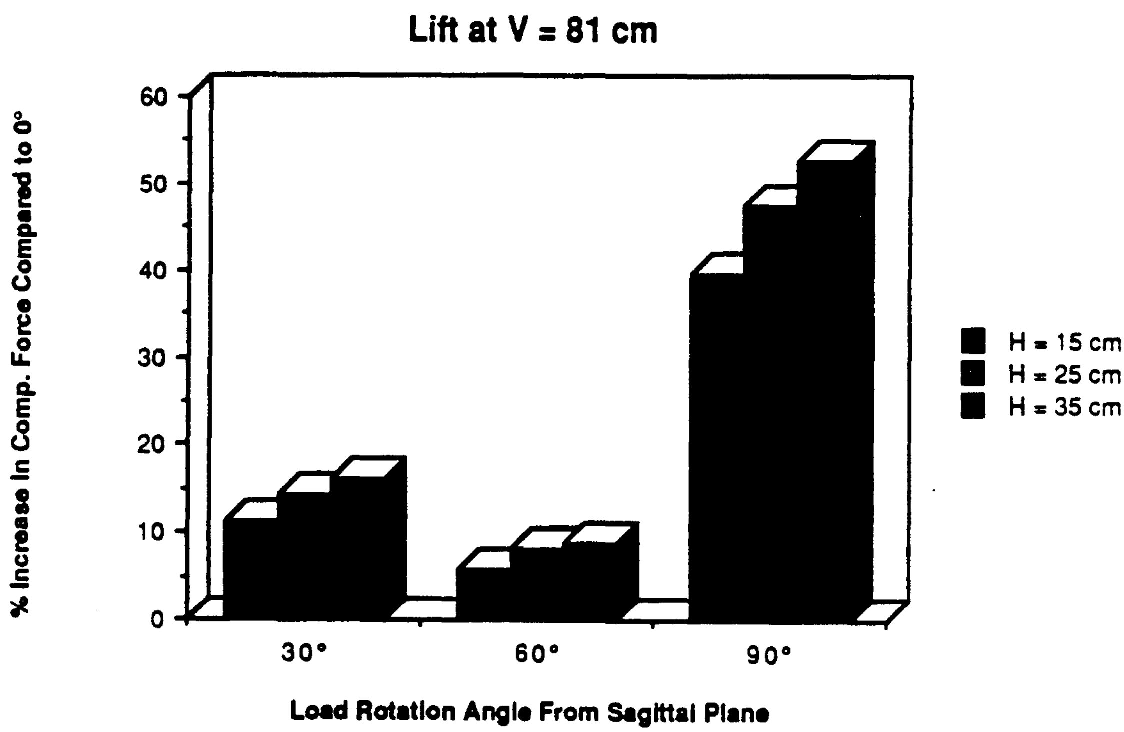


Figure 3.16: Effect of axial rotation of spine on compressive force at L_4/L_5 disc from Chaffin (personal communications).

Predictions of trunk muscle contraction and shear and compressive forces to some extent depend upon the choice of objective function and the assumptions made in the model. For example, Bean et al. (1988) showed that for a given task there was as much as 21% difference in the model predicted compressive force due to differences in objective function.

Maximum voluntary muscle strengths for asymmetric postures have also been studied. Zetterberg et al. (1987) reported significantly higher muscle moments (muscle strength) for trunk flexion and extension than for lateral flexion. Warwick et al. (1980), Garg and Badger (1986) and Garg and Banaag (1988) reported significant and consistent decrease in static lifting strengths with an increase in axial rotation of the trunk (Figure 3.17).

In summary, it appears that asymmetric lifting can produce large and concentrated stresses on lumbar trunk structures. Asymmetric lifting can cause poor postural stability; increased and asymmetric muscle activity; and increased stresses on annulus fibrosus, ligaments and facet joints. Asymmetry in muscle activity can lead to unequal stress concentrations on different component structures of the lumbar spine. It also results in reduced maximum voluntary muscle strength and, therefore, in addition to low-back injury, risk for an overexertion injury is high. Also, injury is more likely in a muscle that is stretched due to twisting or lateral bending and then heavily contracted.

EFFECT OF TASK VARIABLES ON STRESSES TO SPINE

The "safe" weight of the load depends upon a number of task variables and work characteristics. From a biomechanical point of view, some of the task variables that determine "safe" weight of the load include size of the load, horizontal and vertical locations of the load, body posture, foot traction, adequacy of grip, stability of load (solid vs. liquid), obstructions, space constraints, speed of lift, frequency of lift, and direction of force exertion (straight vertical lift vs. pulling the load towards the body) etc. Biomechanical considerations in determining handle positions and angles are discussed in Drury and Deeb (1986). Foot traction has received attention primarily in pushing and pulling tasks (Fox, 1967; Kroemer, 1971; National Safety News, 1974; Lee, 1982). Effects of frequency (fatigue fractures) and asymmetric lifting postures were discussed earlier. Biomechanical effects of horizontal and vertical locations of the load, speed of lifting and lifting technique are briefly discussed here.

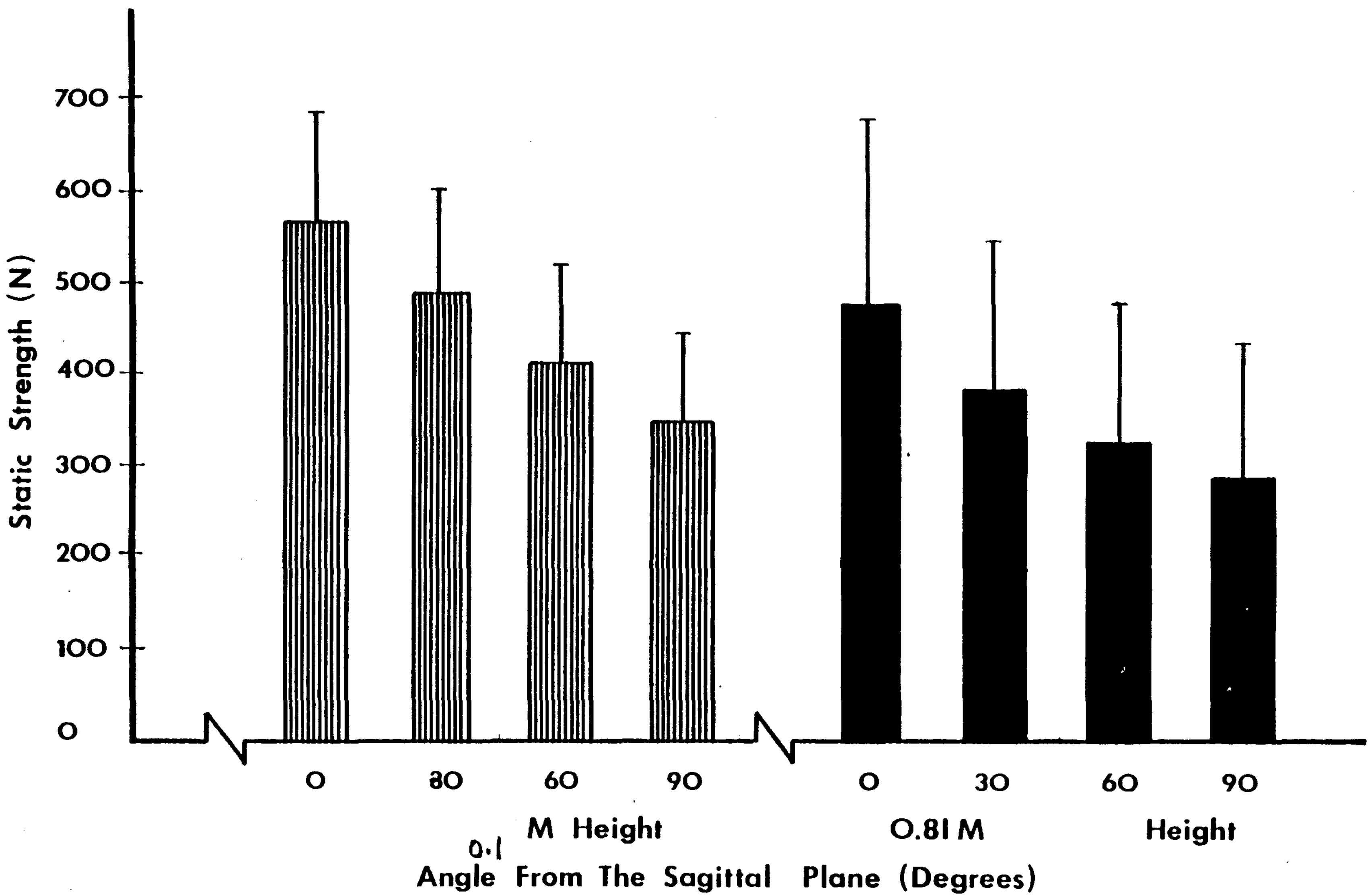


Figure 3.17: Effect of asymmetric lifting on maximum voluntary static strength. (Garg and Banaag, 1988)

Vertical Location of Load

The flexion of the trunk increases as the vertical height of the load to be lifted is lowered from about elbow height. In general, the lower the height of the object to be lifted, the greater the trunk flexion. As the trunk inclines forward, the center of gravity of the upper limbs and torso moves more anterior to the lumbar spine, i.e., moment arm for the upper limbs and torso weight increases (Figure 3.18). This produces larger flexion moment on intervertebral joints. Muscle tension in trunk extensors increases to counterbalance the increased flexion moment. The net result is that the compressive force on lumbar intervertebral discs increases because it is the tension in the back muscles that accounts for a large proportion of spine compressive force (60% when holding a 222 N load; Bean et al., 1988). Secondly, when an object to be lifted is placed on the floor or at a lower height, in most cases it is difficult to bring the object close to the body unless one can straddle the load (keep it between the knees). Once, again, this results in a larger moment arm of the load and, therefore, in higher tension in back muscles and compressive force on the lumbar spine.

In flexed postures, the anterior annulus is under compression, where as the posterior annulus is under tension. Hutton and Adams (1982) and Hickey and Hukins (1980) suggest that the posterior annulus can be severely stretched in flexed postures and this may cause a prolapse of the intervertebral disc. The posterior annulus is also under tension through the circumferential stress generated by the compressive force on the disc (Nachemson and Morris, 1964; Adams et al., 1980). As stated in Nachemson and Morris (1964), approximately 90% of all ruptures in the lumbar discs occur in the posterior part of the annulus. Presence of nerve fibers in the outer part of the annulus can contribute to the production of low-back pain, especially in persons whom the annulus is weakened (Nachemson and Morris, 1964). Also, the rupture of the fibers of annulus could speed the degenerative process. In addition to the disc, the stresses on the posterior ligaments may be fairly high especially in hyperflexion. For example, Adams et al. (1980) reported that supraspinous and interspinous ligaments sprained first in hyperflexion because they were rather weak (tensile strength = 95 N cm^{-2}).

It is believed that a combination of trunk flexion and rotation may be more hazardous to the low back than either one of them alone. Brown et al. (1957) proposed that the disc might be sensitive to torsion and bending. These body motions can cause failure of the annulus fibrosus and hence protrusion of the nucleus pulposus (Hickey and Hukins, 1980). In this regard, Roaf (1960) believes that "a combination of rotation and compression can produce almost every variety of spinal injury."

Several studies have shown an increase in myoelectric activity of trunk muscles with an increase in trunk flexion, until full flexion is reached (Floyd and Silver, 1955; Portnoy and Morrin, 1956; Carlsoo, 1961; Morris et al., 1962; Ortengren and Andersson, 1977; Andersson and Ortengren, 1977). Ortengren et al. (1981) reported that the intradiscal pressure increased when the distance between the floor and the handle height decreased, i.e., when the angle of forward flexion increased. Nachemson (1965, 1980), based on measured intradiscal pressure, reported that the approximate load on the L_3 disc was 686 N when standing erect and 1176 N when the trunk was flexed forward by as little as 20°. Andersson et al. (1977a, 1977b) found a linear increase in intradiscal pressure, myoelectric activity and intra-abdominal pressure with an increase in trunk flexion between 10° and 50° (Figures 3.19 and 3.20).

Thus, a number of indicators of low-back stress such as compressive force (Chaffin, 1969), intradiscal pressure, myoelectric activity and intra-abdominal pressure suggest that lifting at the floor level or at lower heights can be hazardous to the low back and should be avoided through proper workplace design. On the other hand, maximum voluntary muscles strengths are higher at lower heights than at higher heights (Rodgers, 1985; Chaffin and Andersson, 1984; Garg and Banaag, 1988). It appears that the optimum height for lifting may depend on trade-off between the compressive force and muscle strength. However, epidemiological studies suggest that the risk of a back injury is higher when the loads are lifted near the floor level. In this regard, based on a case-referent study, Punnett et al. (1988) concluded that both mild and severe trunk flexions, trunk twist or lateral bending were significant risk factors for back disorders and the risk increased with trunk flexion lasting more than 10% of the work cycle. The study also concluded that the combination of mild flexion and twisting was particularly hazardous. Similarly, Snook et al. (1978) reported that 78% of lifting injuries were associated with low lifting tasks (below knuckle height, 79 cm) even though only 66% of industrial lifting tasks started below knuckle height.

Horizontal Location of Load

Regarding horizontal location of the load, there is almost universal agreement that the load should be kept as close to the body as possible. It is the flexion moment (the product of the weight and the distance from the spinal axis (moment arm)) rather than the weight of the load that is important in determining stresses on the lumbar spine (Figures 3.21 and 3.22). For example, both 200 N (20.4 kg) weight held at 20 cm from the spine and 100 N (10.2 kg) load held at 40 cm from the spine will produce a flexion moment of 40 Nm. However, the load held farther away from the body will require larger trunk flexion which will contribute to the total flexion moment on the lumbar spine as mentioned earlier. Also, bending forward will

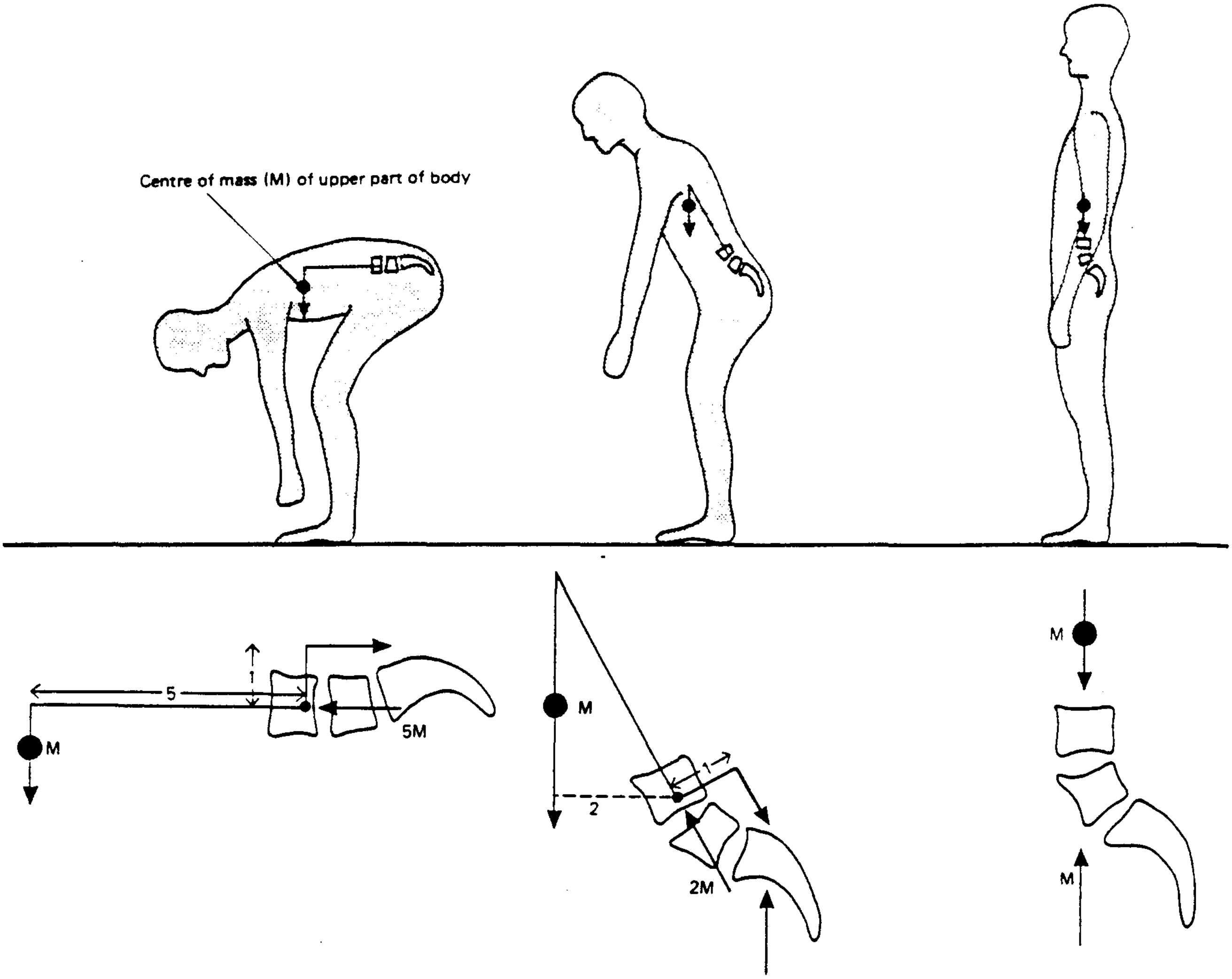


Figure 3.18: Lumbosacral compression equivalent to mass (M) of upper part of body in erect position, increasing as trunk approaches horizontal. (Troup and Edwards, 1985)

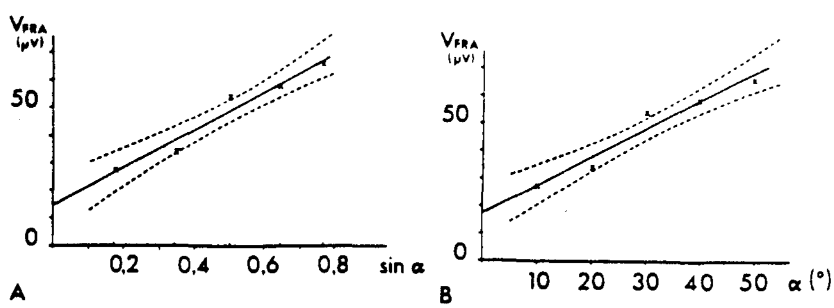


Figure 3.19: Linear regression lines for the relationship between the myoelectric activity at the L3 level, (A) the sine of the angle of forward flexion, and (B) the angle of flexion. Ninety-five per cent confidence regions are indicated. (Andersson and Ortengren, 1977)

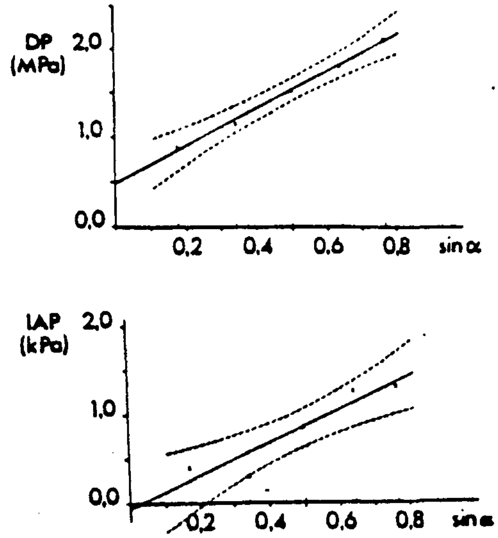


Figure 3.20: Regression analysis of relation between (A, top) disk pressure and the sine of the angle of flexion, and (B, bottom) intra-abdominal pressure and the sine of the angle of flexion. The mean values are denoted with x. The dashed lines indicate 95% confidence limits for the mean values. (Andersson and Örtengren, 1977)

change the curvature of the lumbar spine from lordosis (concave) to kyphosis (convex). Kyphotic posture is generally believed to be undesirable and results in unequal distribution of stresses on the lumbar disc (Grandjean, 1988). It appears that, as far as the low-back stresses are concerned, the flexion moment is probably the most important variable among the various task factors mentioned earlier. Indeed, Schultz et al. (1982) state that, "It can be more stressful to the spine to bend and reach in order to pick up a piece of paper than to lift a large weight held close to the trunk. This is confirmed both by myoelectric activity and the intradiscal pressure measurements."

Chaffin and Andersson (1984) showed that the predicted compression forces on the L_5/S_1 disc increased with an increase in both the weight of the load and horizontal distance of the hands from L_5/S_1 disc (Figure 3.22). Figure 3.22 reveals that an increase of 10 cm in horizontal distance amounts to about a 1000 N increase in compressive force on L_5/S_1 disc. This figure confirms the empirical rule which recommends lifting a load as close to the trunk as possible. Figure 3.23 illustrates the combinations of weight of the load and the horizontal distance from the ankles that would produce 3.43 kN (350 kg) compressive force on the L_5/S_1 disc of the average female at 50 cm vertical height from the floor. The same

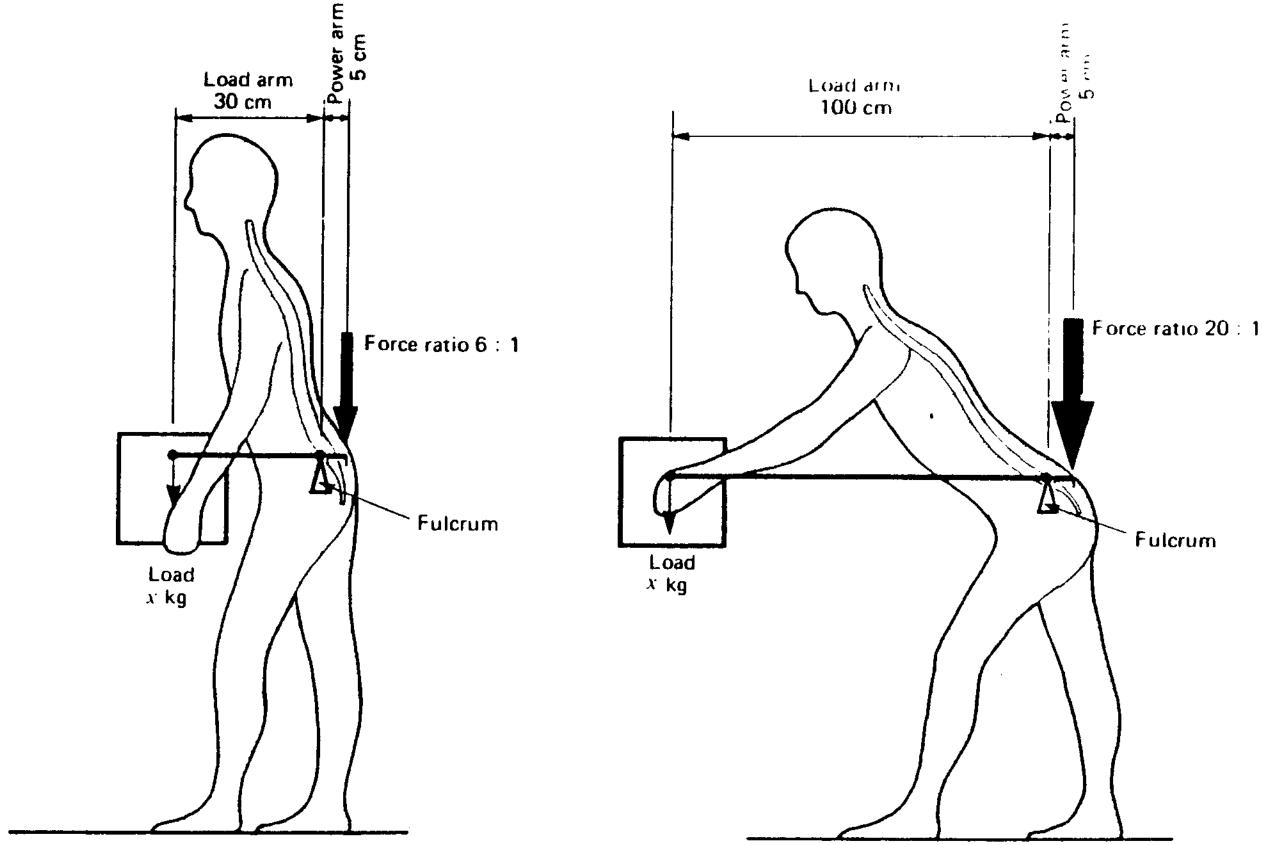


Figure 3.21: Fulcra and levers: increased force is needed to lift a given load when it is further from the body. (Troup and Edward, 1985)

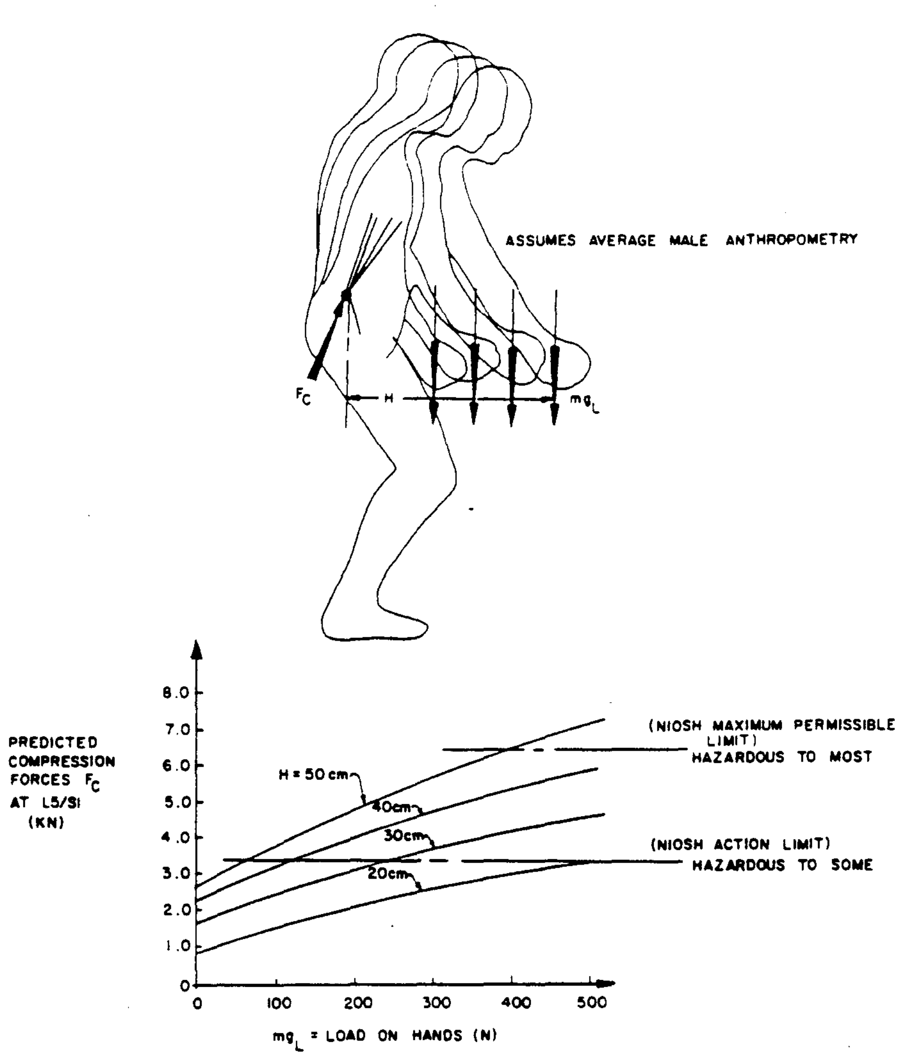


Figure 3.22: Predicted L_5/S_1 disc compression forces for varying loads lifted in four different positions from body. (Chaffin and Andersson, 1984)

combinations which would produce 6.37 kN (650 kg) compressive force on the L_5/S_1 disc of the average male are also shown in Figure 3.23. The compressive force predictions are based on a static biomechanical model (Garg and Chaffin, 1975) and the horizontal location is measured forward of the body centerline from the midpoint between the ankles.

Based on a study of three different warehouses, Garg (1986) observed that the horizontal moments arms were fairly large due to poor workplace design and this resulted in large predicted compressive forces on the L_5/S_1 disc. Based on a laboratory study, Garg (1989) reported that the untrained subjects did not

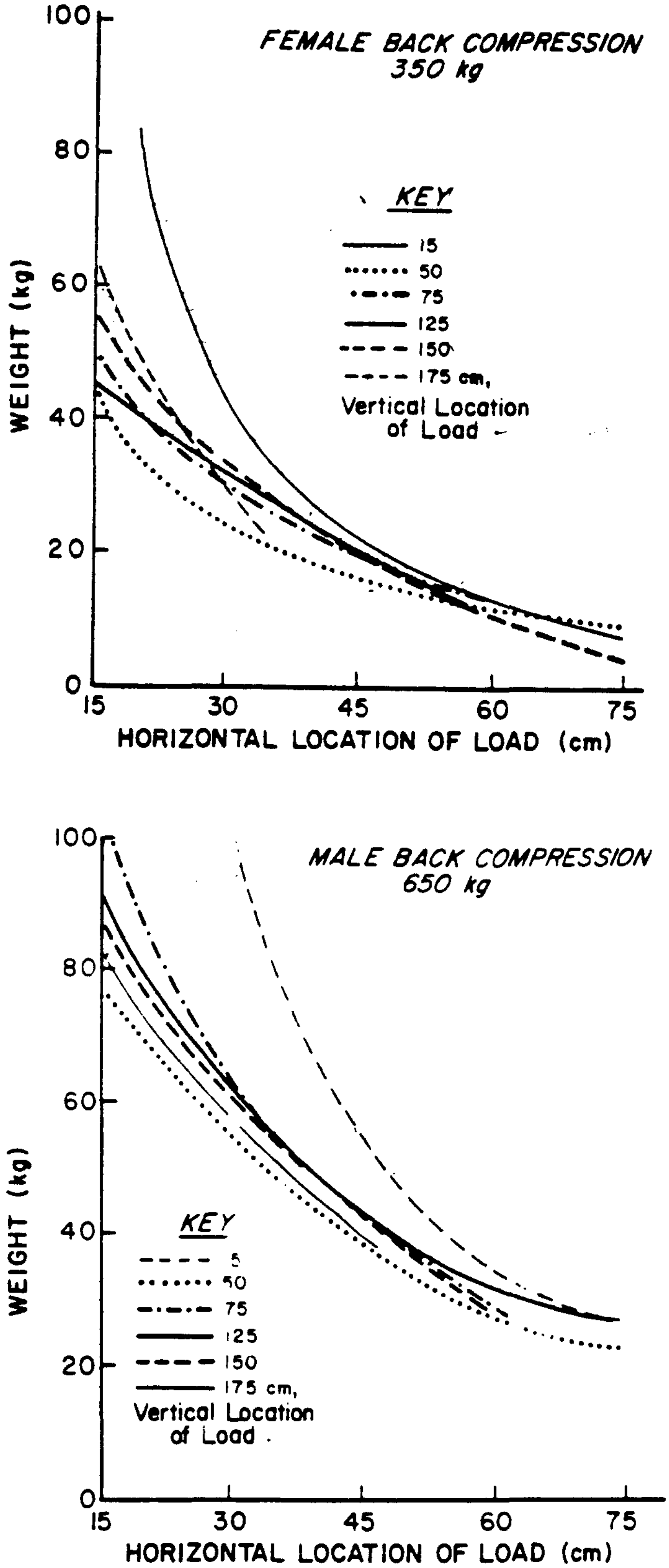


Figure 3.23: Task variables producing 350 kg female (top) and 650 kg male (bottom) back compressions. (NIOSH, 1981)

stand close to the object when lifting moderate to heavy loads from both the floor level and 0.8 m height. Moreover, there was large variation in the horizontal distance between the subjects. Therefore, the same lifting task would have produced different flexion moments and compressive forces on different subjects. Also, workers tend to avoid cardiovascular stresses at the expense of an increased musculoskeletal stress by lifting and carrying greater weight fewer times (McGill and Norman, 1985; Garg et al., 1983; Garg and Saxena, 1985). These issues need to be addressed through proper workplace design, proper work practices, education and training of workers.

Speed of Lifting

Grieve (1970) suggested jerking up a load with sufficient velocity at an early stage to give it energy to take it past the individual's weaker lifting levels might be advantageous, provided the compressive forces are within acceptable limits. As an example, Olympic weight lifters depend on speed and skill to avoid becoming stuck during the lift. This is sometimes justified considering the inverse relationship between the failure strength of the spine and the duration of the applied stress. Also, Gracovetsky (1986) pointed out that the load transmission capacities of the disc are time dependent. However, high speed lifting and, especially, jerking motions cannot be recommended based on the present knowledge of low-back biomechanics. It is quite possible that the inertial forces produced in jerking a load or with high speed lifting can result in momentary but potentially injurious overloads on low-back structures as mentioned earlier under dynamic factors. Sudden bending, twisting and stretching may easily tear the capsular ligaments of the facet joints (Fiorini and McCammond, 1976). With fast motions the ability of the somatic nervous system to coordinate the many muscles necessary to stabilize the spinal column is stressed. Some low-back problems are related to muscle fatigue (Brown, 1973), which further inhibits coordination of the back muscles. Also, high inertial forces associated with dynamic actions are difficult for one to control and a person may lose body balance.

Garg et al. (1982) showed that even at normal lifting speed, inertial forces increased the compressive forces on the L_5/S_1 disc considerably. The peak compressive forces were below 3.4 kN under static conditions and over 6.5 kN at normal speed. Davis et al. (1965) observed that heavier loads were lifted at slower speed. In the initial phases of a lift, inertia of weight of the load, trunk and upper limbs is overcome and acceleration is at its maximum resulting in peak compressive forces (Davis et al., 1965; Garg et al., 1982). Also, at the beginning of the lift the erector spinae muscles remain relatively silent (Floyd and Silver, 1955; Morris et al., 1962) and extension of lumbar spine is delayed, probably to protect against higher peak compressive forces (Davis et al., 1965).

Bush-Joseph et al. (1988) reported that peak moments at the L_5/S_1 joint increased with an increase in lifting speed. They recommended that excessive speed of lifting, including jerking, should be avoided. Marras et al. (1985) reported a significant decrease in torque producing capability of the trunk muscles with an increase in velocity of exertion. For example, there was an average decrease in torque production capability of 35% at only 33% of maximum velocity. Similarly, Kumar et al. (1988) reported an inverse relationship between the peak dynamic strength and the speed of motion. Therefore, it is recommended that the workers should be required to lift loads in a controlled, smooth and well-planned manner. Further, as part of this concern is the need to provide good foot traction and hand grips on such loads to avoid any possible slips and/or falls.

Lifting Straight Up

Based on chrono-cyclophotographic studies, Garg et al. (1983) and Freivalds et al. (1984) showed that the subjects pulled the loads towards the body rather than lifting them straight up vertically. A biomechanical analysis showed that the primary effect of pulling the load towards the body is to transfer stresses from the arms and torso to the legs, which have stronger muscles. Pulling the load towards the body also reduces the compressive force on the L_5/S_1 disc (Garg et al., 1983). A decrease in strength has also been reported when lifting a load straight up vertically (Garg, 1989B; Snook (Personal Communications)). This is a major concern on those jobs where a lift must be performed straight up due to obstructions, workplace constraints or inadequate room.

Lifting Technique

There is no 'natural' way of lifting which is universal. Workers typically use a combination of trunk and knee flexions when lifting an object from the floor. The degree of trunk vs. knee flexion varies from worker to worker and from object to object. Garg and Saxena (1985), based on observations of workers in three different warehouses where they were required to lift loads frequently, found that practically no workers bent their knees and kept their backs straight when lifting objects near the floor level. Other studies have also observed that the stoop lifting technique is a frequently used technique (Brown, 1971; Park, 1973; Shephard, 1974; Stubbs, 1976; Troup, 1979). Davis et al. (1965) reported that young adult untrained males when asked to lift a heavy weight by using their legs and not their backs tended to convert the bent knee lift into a straight knee lift. Many authors and most physicians and safety professionals recommend that a squat posture (straight-back, bent-knees) be assumed when lifting a load from the floor level (Asmussen et al., 1965; Adams and Hutton, 1982; Leskinen et al., 1983; Ayoub and El-Bassoussi, 1978). Squat lifting technique is said to be preferable to stoop lifting because (i) the load is closer to the body resulting in smaller moment arm (Bendix and Eid, 1983; Troup et al., 1983), (ii) it shifts the load to the legs which are stronger than the back, (iii) the low-back ligaments are exposed to a lower maximal strain (Poulsen, 1981; Anderson, 1983), (iv) it produces relatively smaller compressive force on the lumbar back (Ayoub and El-Bassoussi, 1978; Leskinen et al., 1983), and (v) the spine is kept in the sagittal plane, i.e., neither rotated nor bent sideways (Troup and Edwards, 1985). In considering the biomechanical implications of stoop vs. squat postures, often it has been assumed that the load can be brought between the knees with the squat posture and it is farther away from the body with the stoop posture (Figure 3.24). This is not always true. For example, Garg et al. (1983) reported that the horizontal distances (from ankles to the hand grips) were a little lower for the free style (stoop) postures than those for the squat postures for lifting boxes ranging in width from 25 cm to 63 cm. Biomechanical evaluations have shown that stoop lifting method is preferable in those

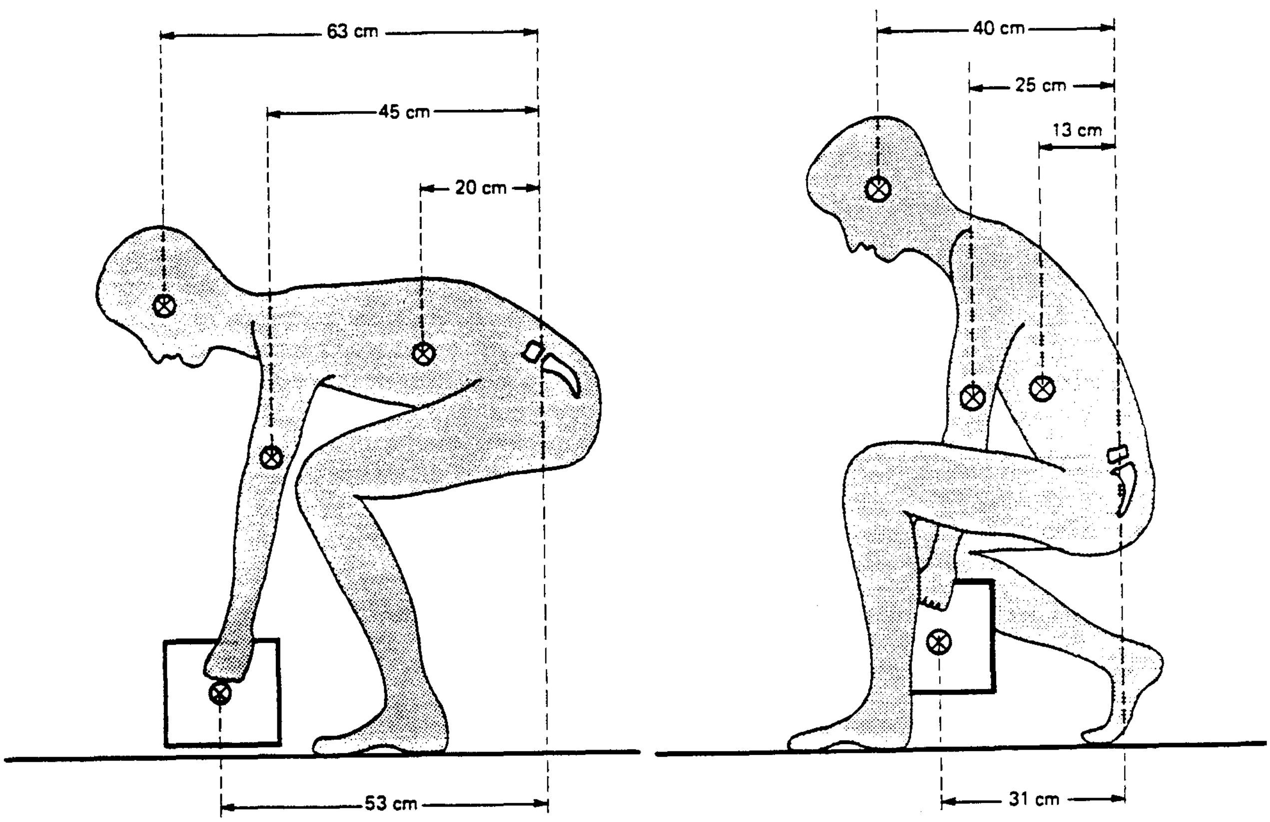


Figure 3.24: Vertical lifts from stooping position with load in front of knees (left) and from crouching posture with load between knees (right). The circled crosses indicate the centers of mass for the head, upper limbs, trunk and load: the horizontal distances from the lumbosacral disc are shown. (Troup and Edwards, 1985)

situations where the load is too wide or bulky to be placed between the knees (Park and Chaffin, 1974; Garg and Herrin, 1979; Frankel and Nordin, 1980; Troup et al., 1983). In these situations, the stoop lifting posture is preferable (Figure 3.25) because (i) the load moment arm may be smaller and (ii) the vertical components of both the body weight and load are shifted to a shearing effect when the spine is bent (Chaffin, 1975; Garg and Herrin, 1979). Of course, the shearing forces on the L_5/S_1 disc are greater when lifting with the bent spine and the articular facets and the annulus fibrosus provide the

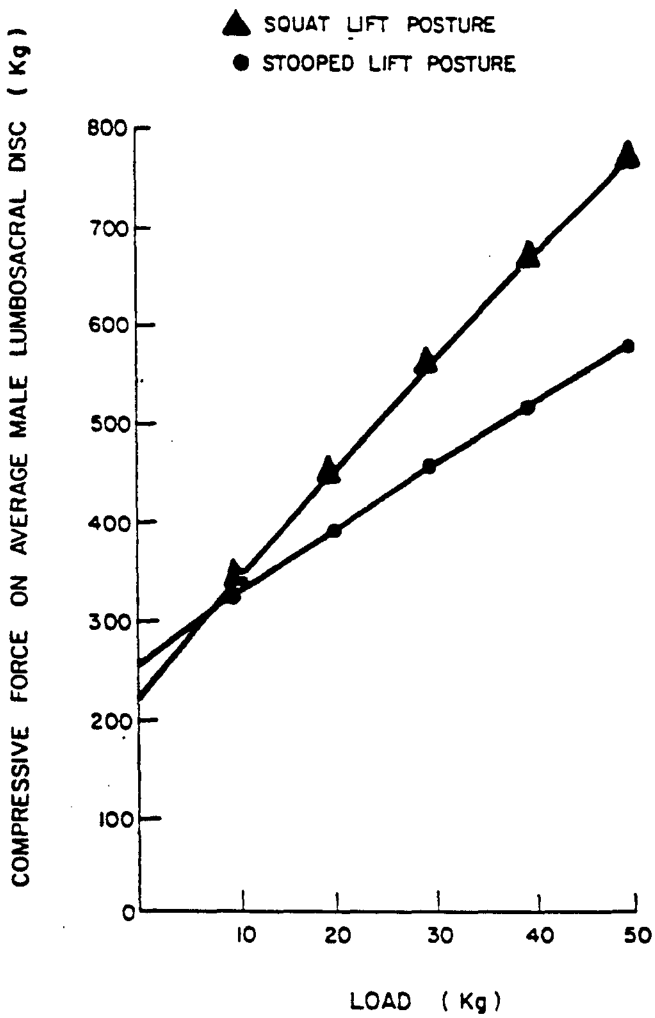


Figure 3.25: Effect of load on L_5/S_1 compressive force for large load in front of body. (Garg and Herrin, 1979)

primary resistance to these forces. Other disadvantages of squat lifting postures are that (i) most workers' knee extensor muscles are a weak link in the lifting process (Troup and Edwards, 1985; Bejjani et al., 1984; Davis et al., 1965), (ii) the feet usually are not flat on the floor and may not be far enough apart for good stability and leverage, and (iii) the lifting strength is lower as compared to that in stoop posture (Garg et al., 1983; Marras et al., 1984). As mentioned earlier, most workers use a combination of trunk and knee flexions. This does not imply a hyperflexed back, which should be avoided as it places greater stress on the posterior ligaments and posterior annulus of the disc. Soft tissues of the trunk can be damaged if the spine were hyperflexed (Hutton and Adams, 1982).

Nachemson and Elfström (1970), based on intradiscal pressure measurements, reported compressive forces of 3330 N and 2050 N for bent-back, straight-knee and bent-knee, straight-back postures, respectively, for lifting a 20 kg load. However, Ortengren et al. (1981) found no significant difference in intradiscal pressures for the two methods of lifting.

Some authors and most back schools emphasize the importance of keeping the back flat throughout the lift (Asmussen et al., 1965; Grandjean, 1988; Anderson and Chaffin, 1986). A flat back is one in which the lumbar lordosis is maintained throughout the lift by holding the trunk rigid and forcing all torso flexion to occur at the hip joints. The flat-back configuration is recommended because it provides the greatest amount of muscle control of the trunk and minimizes strain on posterior spinal ligaments. However, others argue that the flat back would prevent midline ligament from contributing to the reduction in lumbar stresses (Farfan, 1975; Gracovetsky et al., 1981; Gracovetsky et al., 1985). These authors suggest that some flexion of the spine is advantageous as it would transfer some of the forces to the posterior ligaments. These authors recommend flexing the spine and bending the knees as it would result in the lowest possible compression stress on the spine. This recommendation is also supported by Hutton and Adams (1982), who reported that wedged lumbar intervertebral joints to simulate forward flexion resulted in greater compressive strength. These authors recommend flexing the lumbar spine and extending the thoracic spine to keep the load close to the body. Adams and Hutton (1982) recommend that it is mechanically and nutritionally advantageous to flex the lumbar spine when lifting heavy weights because it reduces the compressive stress on the posterior annulus and improves transport of metabolites in the intervertebral discs. On the other hand, based on a biomechanical evaluation of five different lifting methods, Anderson and Chaffin (1986) reported that, in general, the straddle stance with flat back produced the lowest compressive force on the L_5/S_1 disc when lifting both compact and bulky loads. They also reported that lifting methods incorporating a flat back resulted in minimum risk to lumbodorsal fascia (%

strain) and the methods involving a curved back in the maximum risk. Anderson and Chaffin (1986) concluded that a straddle-stance with a squat lift should be used so that even the bulky load could be brought close to the body and the back should be kept flat (as when standing erect).

Thus, there are differences between various techniques in terms of flat vs. curved back, bent knees vs. straight knees, placement of feet and placement of hands. Different lifting techniques may produce different compressive forces, intradiscal pressures and ligament stresses and moment producing capacities of the muscles may also be affected. At present biomechanical advantages of one lifting technique over others are not clearly established due to controversy in the literature cited. Protection of the other body segments, such as knee joints in addition to the back, must also be considered and one must maintain stability over the entire lifting activity. More research is necessary to establish the validity of any suggested method of handling loads. However, all lifting techniques agree that the load should be kept as close to the body as possible and the load should be lifted in a slow and controlled manner.

CONCLUSIONS

(1) Lifting compact objects of moderate weights close to the body could create compressive forces sufficient to cause damage to some lumbar intervertebral discs, especially to the lumbar spines of older workers. For example, lifting an object weighing less than 245 N (25 kg) within 25 cm from the ankles can create compressive forces in excess of 3.43 kN (350 kg).

(2) The compressive force is directly related to the horizontal moment arm of the load. Therefore, even light loads should be lifted close to the body through proper workplace design and education and training of workers.

(3) Both the intradiscal pressure and compressive force increase with an increase in angle of trunk flexion. When a load is lifted from the floor stresses on the low back increase due to larger moment arms of the body weight and weight of the load. Therefore, objects, especially heavy loads, should not be stored on the floor, but should be raised to about standing knuckle height (minimum 50 cm) to avoid the necessity for stooping over and lifting.

(4) Asymmetric lifting (with one hand or with both hands at the side or with torso twisted and/or laterally bent) results in significantly lower maximum voluntary muscle strength and higher compressive force, intradiscal pressure, myoelectric activity and antagonistic activity of trunk muscles. These postures can impart complex and potentially hazardous stresses to the lumbar column. Asymmetric

lifting should be avoided by proper workplace design, proper work practices, and education and training of the workers.

(5) Frequent lifting of moderate to heavy loads can lead to fatigue fracture of the lumbar intervertebral joints, in addition to cardiovascular stresses and localized muscle fatigue. Thus, repetitive lifting may cause disc degeneration in addition to low-back problems due to muscle fatigue.

(6) Rapid and/or jerking motions can impart substantially higher and potentially dangerous stresses to the low-back structures. Maximum voluntary muscle strength also decreases with an increase in speed of lifting motion. Workers should be required to lift loads in a smooth and controlled manner.

(7) Practically all lifting studies have been conducted with good foot traction. Weights should be reduced significantly and the objects should be lifted in a planned and careful manner when a low coefficient-of-friction exists in a workplace. Slips and falls can cause serious back injuries.

(8) The different lifting techniques recommended to lift loads from the floor can cause different complex stresses on different parts of the low-back structure (muscles, ligaments, disc and facet joints). Effects of different lifting techniques on stresses to low back and other parts of the body are not fully understood. Further, a safe lifting technique may depend upon such factors as leg strength, weight of the load, size and shape of the load and workplace geometry. Until such complexities are better understood, it is recommended that instructions on lifting techniques be avoided.

(9) Jobs should be designed so that the compressive forces on the L_5/S_1 disc are below 3.43 kN (350 kg) to protect most workers. This upper limit will not be protective for older workers and for those with weaker spines. Also, jobs which place more than 6.4 kN (650 kg) of compressive force on low back are hazardous to all but the healthiest of workers.

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