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Workplace Risk Factors and Occupational Musculoskeletal Disorders, Part 1: A Review of Biomechanical and Psychophysical Research on Risk Factors Associated with Low-Back Pain

W. Monroe Keyserling [a]
[a] Department of Industrial and Operations Engineering, Center for Ergonomics, The University of Michigan, 1205 Beal Ave., Ann Arbor, MI 48105-2117.

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AUTHOR
W. Monroe Keyserling

Department of Industrial and Operations Engineering, Center for Ergonomics, The University of Michigan, 1205 Beal Ave., Ann Arbor, MI 48105-2117

Workplace Risk Factors and Occupational Musculoskeletal Disorders, Part 1: A Review of Biomechanical and Psychophysical Research on Risk Factors Associated with Low-Back Pain

Injuries and disorders caused by overexertion and repetitive motion are the leading causes of compensable lost-time cases in the United States. Epidemiological and laboratory-based research methods have been used to evaluate the significance of various risk factors associated with overuse injuries and disorders. The National Institute for Occupational Safety and Health performed a comprehensive review of over 600 epidemiological studies in 1997 and concluded that there was evidence of a causal relationship between low-back injuries and disorders and workplace exposures to forceful exertions, awkward posture, and vibration. Although epidemiological studies provide important insights to understanding the causes of work-related overuse disorders, they are sometimes criticized for their inability to measure precisely how people respond to specific risk factors found in the workplace. This article presents a review of recent laboratory studies and biomechanical models of work factors believed to be associated with increased risk of low-back injuries and disorders. Biomechanical models and laboratory studies do not replace epidemiological studies. However, these approaches provide important complementary information that is needed to understand the complex process of how exposures to physical risk factors result in strain that may ultimately lead to injury or disease. These studies also provide important insights as to how people react and respond to specific physical risk factors found in the workplace. Combined with epidemiological research, laboratory studies are an essential element in understanding the causes and prevention of work-related overexertion injuries.

Keywords: biomechanics, ergonomics, low-back pain, psychophysics

Data collected through the U.S. Department of Labor's Annual Survey of Occupational Injuries and Illnesses (ASOII) demonstrate the high morbidity of work-related overexertion injuries and disorders in the United States. A recent monograph published by the National Institute for Occupational Safety and Health (NIOSH) summarizes several key findings of the 1994 ASOII:

- For injury and illness cases involving days away from work, approximately 706,000 cases (32% of the total) resulted from overexertion or repetitive motion.
- Approximately 530,000 lost work-time cases were associated with manual materials handling activities such as lifting, pushing, pulling, and carrying. The low back was the most common site of injury associated with materials handling

activities; over 60% of these cases involved back pain. Injuries to the shoulders accounted for approximately 13% of materials handling cases. The median time away from work for materials handling cases was 6 days.

- Approximately 93,000 lost work-time cases resulted from repetitive motion, such as data entry tasks, repetitive use of tools, and repetitive manual transfer tasks. The median time away from work associated with these cases was 18 days.
- The remaining 83,000 lost work-time cases were associated with unspecified overexertion events.

The true cost of work-related overexertion injuries and disorders in the United States is not known. Conservative estimates of annual expenditures, based on workers' compensation payments (indemnity and medical services) and other direct costs, range between $13–20 billion.(1) The total cost to society is believed to be substantially higher due to various indirect costs (e.g., lost productivity, costs of hiring and training replacement workers, overtime, administrative costs, and miscellaneous transfer payments) that are not included in the conservative estimates. The total annual societal cost has been estimated to be as high as $100 billion.(2)

Epidemiological and laboratory-based research methods have both been used to evaluate the significance of various risk factors associated with work-related musculoskeletal disorders. Epidemiological studies are designed to look for significant associations between exposure to ergonomic risk factors (e.g., force, repetition, posture) and selected health outcomes (ranging from medically diagnosed disease entities to subjective reports of pain or discomfort) in selected populations of workers. NIOSH(1) performed a comprehensive review of over 600 epidemiological studies of occupational musculoskeletal disorders. This study concluded that there was either "strong evidence" or "evidence" of a causal relationship between workplace exposures to forceful exertions, repetition, and awkward posture and musculoskeletal disorders of the neck, upper extremity, and low back. This review also found "strong evidence" of a causal relationship between low-back pain and whole body vibration, and between segmental vibration and hand-arm vibration syndrome. Epidemiological studies provide important insights to understanding the causes of work-related musculoskeletal disorders. However, these studies are sometimes criticized due to their inability to precisely measure exposures to risk factors and the associated biomechanical and/or physiological responses to these exposures.

Biomechanical models and laboratory studies do not replace epidemiological studies. However, these approaches provide important complementary information in the quest for understanding of the complex process of how exposures to ergonomic risk factors result in physiological responses that may ultimately lead to work-related injuries and illnesses. This article presents a review of laboratory studies and biomechanical models of risk factors associated with low-back pain in the NIOSH(1) epidemiological review. The accompanying article(3) addresses laboratory studies and biomechanical models of work factors associated with increased risk of upper extremity injuries and disorders.

Laboratory studies are controlled scientific investigations of how humans respond when exposed to specific ergonomic risk factors (e.g., forceful exertions, awkward work postures, high repetition, etc.) during simulated work activities. Responses include both objective biomechanical/physiological measurements such as the electromyographic activity of a working muscle and subjective psychophysical measurements such as ratings of perceived exertion. Most of the studies cited below and in the accompanying article were performed in true laboratory settings. A few studies were performed in operational work environments modified as necessary to collect data under carefully controlled conditions.

Because of ethical issues related to the protection and safety of human subjects, laboratory studies are designed to keep exposures to risk factors at levels below the threshold of injury. As a result, these studies are generally incapable of "proving" a relationship between exposure and injury. Despite this limitation, laboratory studies provide important scientific insights as to how the body responds to ergonomic stresses. Combined with pathophysiological models of musculoskeletal injury mechanisms and epidemiological findings of positive relationships between exposure to ergonomic risk factors and musculoskeletal injury, laboratory studies are an essential element in understanding the causes and prevention of work-related overexertion injuries.

This article also reviews biomechanical models that simulate and/or predict how the musculoskeletal system responds to work factors such as work posture, external loads placed on the hands, and movement dynamics. These models can be used to estimate musculoskeletal stresses in the absence of a human experiment.

BIOMECHANICAL APPROACHES TO STUDYING ERGONOMIC RISK FACTORS

The discipline of occupational biomechanics is concerned with measuring and/or modeling the "internal" mechanical responses of body tissues to the "external" physical demands of a work activity. These external demands include (1) the magnitude and direction of force(s) exerted while working (e.g., weights lifted during manual handling tasks, exertions required to operate tools and equipment), (2) the location(s) where the external force acts on the body, (3) the posture(s) required to perform the job, and (4) movement dynamics (velocity and acceleration). A variety of methods have been developed to measure or predict internal responses to external demands, including the following.

- Direct measurement of a specific response, such as using small transducers to measure internal pressures in the spinal disks during lifting or in the wrist during hand-intensive work.(4–6)
- Indirect measurement of a specific response, such as using electromyography (EMG) to determine the magnitude of force exerted by a muscle group when performing a task. When a muscle fiber twitches, a small electrical potential is generated that can be measured by placing an electrode either within the muscle or on the surface of the skin above the muscle. (Surface electrodes are less invasive and are generally used in laboratory and workplace experiments.) Because the relationship between the force exerted by a muscle and the electrical output of the muscle is monotonic, calibration procedures are used to convert EMG signals into estimates of force output.(7,8) Since it is impractical to measure directly the magnitude of muscle exertions, EMG provides a reasonable approach for estimating forces exerted during work activities.(9)
- Mathematical modeling to predict how the body reacts to the physical demands of a job, such as using kinematic models of the musculoskeletal system to estimate compressive forces on spinal disks, strength requirements of "whole body" exertions, or tension/compression stresses on wrist tendons.(10) In addition to estimating responses to specific work requirements, these models can also be used to evaluate alternative work situations and to select configurations of layout, tooling, and job demands that do not place excessive stresses on musculoskeletal tissue.

Biomechanical methods have been used extensively in both laboratory and field settings. In some instances, direct and indirect measurement methods have been used to confirm the predictions of models.

PSYCHOPHYSICAL APPROACHES TO ERGONOMICS RESEARCH

Psychophysics is a discipline dating back to the 19th century that bridges psychology and physics by examining the relationships between physical stimuli in the environment and the resulting sensations perceived by the humans who are exposed to the stimuli. Various quantitative relationships have been proposed, culminating in the general model proposed by Stevens:[11]

$$S = kI^n$$

where S is the intensity of the perceived sensation, I is the intensity of the physical stimulus, k is a constant (determined by the units of measurement used), and n is an exponent that varies for different types of physical stimuli.

The exponent n describes the strength of the relationship between the intensity of the stimulus and the intensity of sensation. Values of n have been determined empirically for stimuli commonly encountered by humans, such as 3.5 for electric shock, 0.6 for loudness, and 1.6 for muscular effort.[12–14]

In the field of ergonomics, psychophysical methods have been used to empirically determine acceptable levels of work intensity by asking subjects to adjust their work loads (the physical stimulus) so that the resulting discomfort/fatigue (the perceived sensation) is acceptable. Approximately 35 years ago, researchers at the Liberty Mutual Research Center[15] initiated a series of psychophysical experiments to evaluate human responses to common manual materials handling tasks and to develop guidelines for job design. The initial Liberty Mutual study focused on lifting with maximum acceptable weight (MAW) being the dependent variable of primary interest. Subsequent studies at Liberty Mutual and other research labs have examined materials handling activities such as carrying, pushing, pulling, and lowering,[16–23] and more recently, repetitive hand motions.[24–26]

The psychophysical approach also has been used to evaluate the stressfulness of work by having subjects in a laboratory or workers on the job rate their perceptions of the intensity of work effort and/or their perceptions of discomfort. Borg[14] demonstrated a strong correlation between heart rate and perceived exertion rated using 10-point and 20-point categorical rating scales. Derivations of the Borg methodology have been used by numerous ergonomic researchers to obtain perceptions of exertion and discomfort in a variety of work and simulated-work situations.[27–31] Results from these studies have been used to recommend modifications to existing workplaces and to develop guidelines for job design.

Table I gives a summary of the key articles reviewed for the present article. Factors associated with an increased risk of work-related low-back pain are listed in the left-most column. For each risk factor, the results of relevant laboratory experiments and biomechanical model predictions are briefly described. A more detailed discussion of the literature is presented in the following sections.

BIOMECHANICAL FACTORS IN WORK-RELATED LOW-BACK PAIN

Low-back pain symptoms are caused by a variety of injuries and disorders. Although the underlying cause of back pain cannot be determined definitively in up to 90% of patients, work-related cases are believed to result from the following mechanisms: (1) muscle or ligamentous injury, (2) herniation of the intervertebral disc with irritation of adjacent nerve roots, and (3) degenerative changes in the intervertebral discs.[32] Significant biomechanical research has been devoted to understanding how the musculoskeletal tissues of the lower back are affected by the parameters of job demands, such as the postures required to perform a job and/or the forces exerted during manual materials handling tasks. These studies have focused on evaluating how work requirements challenge the strength capabilities of muscles and connective tissues, and the load bearing capacities of the spinal motion segments. In these studies the human body is treated as a mechanical system, made up of rigid links (the bones) that are connected at joints. Forces and mechanical moments (torques) imposed on the system during work activities are estimated by static and dynamic biomechanical models and then compared with the strength capabilities and biomechanical tolerance limits of the affected tissues. The job is considered to be potentially hazardous if the imposed forces or moments exceed the strength or biomechanical tolerance limits of either an individual or an agreed-on percentage of the population.[33–35]

Static Whole Body Kinematic Models

Chaffin and associates at The University of Michigan developed a static three-dimensional kinematic model of the musculoskeletal system that can be used to evaluate biomechanical responses to whole body exertions such as lifting, pushing, and pulling.[10,36,37] This model generally has been used for two purposes: (1) to compare the strength demands of a task with the strength capabilities of the workforce to estimate the percentage of adult males and females who are capable of performing the task, and (2) to predict compressive forces acting at the L5/S1 spinal disc during static exertions. The model has been used extensively to evaluate whole body tasks that are performed at normal (nonjerky) movement speeds on an infrequent basis (typically less than once every 5 min). Because the model does not consider the effects of fatigue, it is generally not appropriate for highly repetitive tasks (psychophysical or metabolic job analysis tools are preferred) or highly dynamic motions (dynamic models are preferred). Despite these limitations, the model has been used to predict biomechanical responses to strenuous exertions associated with common manual handling tasks.

Static Biomechanical Analysis of Strength Demands

To use the Michigan model, it is necessary to describe the worker's anthropometry (height and body weight), working posture (angles at the ankles, knees, hip, trunk, shoulders, and elbows), and the vector (magnitude and direction) of the external load acting on the hands. The model uses this information to compute the strength required at the ankles, knees, hip, trunk, shoulders, and elbows to maintain the system in static equilibrium. Individual and task parameters such as body weight, posture, and hand force create resultant forces and mechanical moments at each joint. To maintain equilibrium, each joint must exert an equal and opposite reactive force and the muscles at each joint must have sufficient strength to create an equal and opposite reactive moment. For this reason, strength is characterized as the ability to create a mechanical moment.

The Michigan model has been used to demonstrate how posture affects the strength demands of a job. For example, if a person

TABLE I. Workplace Risk Factors Associated with Low-Back Pain—A Summary of Literature Describing Laboratory Experiments and Biomechanical Models

Workplace Risk Factor	Physiological Measurements	Biomechanical Model Predictions and Biomechanical Tissue Studies	Psychophysical Measurements
Trunk forward flexion	EMG activity in trunk extensors[38–41] Intradiscal pressure[4,40]	Strength demand on trunk extensors)[10, 62] Compression force on spinal disks[10] Shear force on spinal disks[62]	
Trunk rotation (axial twisting)	Asymmetric EMG activity level in left and right erector spinae muscles[63,64] Increase in EMG activity in trunk extensor muscles relative to EMG activity during symmetric lifts[70]	Left/right shear force on spinal disks[63–64] Nonsagittal components of spinal segment motions[69]	Maximum acceptable weight decreases with increased trunk twisting[21,23,72,73]
One-handed lifting		Nonsagittal components of spinal segment motions[69]	Maximum acceptable weight decreases with one-handed lifting[23]
Lifting above shoulder height			Maximum acceptable weight decreases when lifting above shoulder height[19–22]
Lifting in restricted work postures			Maximum acceptable weight decreases under conditions of a restricted ceiling height[23,76]
Magnitude of lifting force (object weight)	Intradiscal pressure[4,40] Electromyographic activity[70]	Strength demand on trunk extensors and upper extremity muscle groups [10,62] Compression force on spinal disks[10] Mechanical failure of cadaver tissues under compressive loads[35,52]	
Horizontal location of center-of-gravity	Intradiscal pressure[4,40]	Strength demand on trunk extensors and upper extremity muscle groups[10,62] Compression force on spinal disks[10] Mechanical failure of cadaver tissues under compressive loads[35,52]	Maximum acceptable weight decreases as load size (displacement of center-of-gravity) increases[19–22,72]
Availability of handles			Maximum acceptable weight greater for objects equipped with handles[74–75]
Task frequency/repetition	Decrease in EMG mean power frequency in spine extensor muscles with increased lift frequency[70]	Reduced threshold for failure of cadaver tissues under cyclical loading[55–57]	Maximum acceptable weight decreases as frequency increases[19–22,73] Maximum acceptable push force decreases as frequency increases [22] Maximum acceptable pull force decreases as frequency increases[22]
Task duration/shift length			Maximum acceptable weight decreases as shift length increases from 8 to 12 hours[19,20]
Distance (displacement of lifted, pushed, or pulled object)			Maximum acceptable weight decreases as distance increases[22,23] Maximum acceptable push force decreases as distance increases[22] Maximum acceptable pull force decreases as distance increases[22]
Dynamic effects of lift velocity (acceleration of external loads and body segments)		Compression force on spinal discs[62–66]	

with average male height and weight stands upright (trunk vertical) with the arms flexed at a 45° angle below the horizontal, the required lower back strength to counteract the resultant mechanical moment is only 14.6 Newton-meters (Nm). If this person bends forward at the hip so that the trunk is at a 45° flexion angle, the required strength increases to 121 Nm.[10] The posture change has displaced the weight of the trunk, head, and upper extremities forward, increasing the moment arm and the required strength at the low back. Placing a load in the hand would increase the strength demands at the low back even further, due to the increased resultant moment created by the external load.

The extensor muscles of the lower back must work harder during trunk flexion to counter the increased mechanical moment. This has been confirmed in numerous laboratory studies using EMG to measure the activity of back muscles during static trunk flexion.[38–41] Andersson et al.[42] found that the EMG activity in the lumbar back muscles increased linearly from a mean of approximately 25 μv at 10° of forward flexion to approximately 60 μv at 60° of flexion.

As already discussed, the Michigan model computes the strength demands imposed by a task on major body joints. The ability (or lack of ability) to perform the task is a function of the strength capacities (i.e., the capability to produce the required reactive mechanical moments) at the joints.[43] When a person attempts to lift, carry, push, or pull, the resultant moments created at each joint due to the load in the hands and the body weight must be less than or equal to the strength at each joint. After the model has computed the strength required at each joint to perform a specific task, this value is compared with statistical distributions of the strength capabilities of U.S. adults to estimate the percentage of males and females who are capable of performing the task.[10] A task with a high percentage capability prediction can be performed by most workers, whereas a task with a low percentage capability prediction exceeds the ability of many workers.

There is limited evidence that strength capability may be related to the risk of overexertion injuries. Maximum isometric lifting strength was compared with job strength demands in a study of 411 workers in an electronics manufacturing facility.[44] The strength tests were designed to replicate job demands. During a 1-year follow-up, the rate of low-back injuries was three times greater in workers who did not demonstrate strength equal to or above job requirements; however, the sample size was not sufficient for statistical significance. In a similar study of aluminum and rubber workers, persons whose maximum isometric strength matched job demands had fewer injuries than an unmatched group; again, however, the results were not statistically significant.[45] A study of aerospace workers found no relationship between isometric strength and back injuries; however, the strength tests in this study were not designed to simulate job tasks.[46]

The injury studies cited in the previous paragraph compared the strength demands of a job with an individual's strength as determined by an isometric test. Strength is only one of many personal and task risk factors associated with low-back pain.[1] The mixed results reported above may be attributed to the fact that the individual worker was the unit of analysis, increasing unexplained variance due to unmeasured personal, workplace, and task variables. In a study designed to reduce variance associated with personal factors, the Michigan model was used to evaluate the relationship between injuries and model-predicted population strength demands on the jobs of 6912 workers in 5 different industries.[47] For jobs where strength demands exceeded the model-predicted strength capability for the weakest 10% of the population, a significant ($p < .05$) increase was observed in reported back pain (sprains, strains, degenerative disc disease, and nonspecific pain). The NIOSH Lifting Equation[33,34,48] considers population strength capability when evaluating the safety of a lifting task. For a job to be classified "acceptable," the strength requirements must be within the strength capability of at least 75% of the female working population. (Note: NIOSH used a combination of biomechanical and psychophysical strength considerations in developing the computational formula for the lifting equation.)

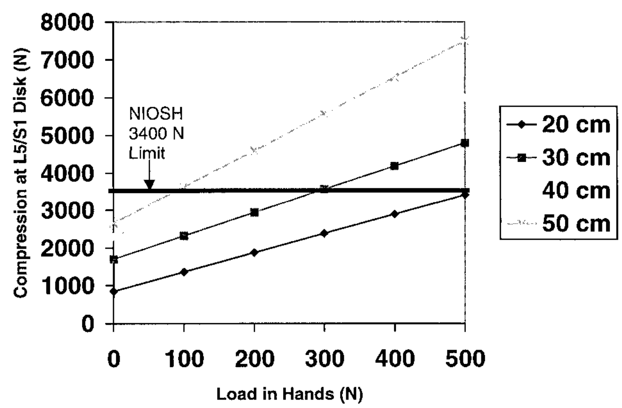

FIGURE 1. Predicted L5/S1 disc compression forces for varying loads lifted at four horizontal displacements from the body (from Chaffin and Andersson, 1991)[10]

Static Biomechanical Analysis of Compression Forces on the Spinal Discs

From a biomechanical perspective, the fact that large resultant moments are induced in the lumbar spine during lifting and/or forward bending raises the question of the nature of internal forces that must be exerted to create the required reactive moments to stabilize the spine. An early biomechanical model developed in 1961[38] assumed that two types of internal forces act to resist the resultant flexion moment. The largest contributor to the stabilizing reactive moment is the erector spinae muscles, located approximately 5 cm posterior to the center of rotation of the spinal discs. Through forceful contractions, these muscles create an extension moment to maintain stability. The second stabilizing force is created by intra-abdominal pressure, which pushes upward on the diaphragm in front of the spinal column, creating a small extension moment to resist the external load. Recent studies[49–50] have questioned the significance of the contribution of intra-abdominal pressure in creating the extension moment, so it is frequently not considered in contemporary models. Another recent study[51] has shown that the moment arm of the erector spinae muscles is displaced approximately 5.5–7.0 cm from the center of rotation of the spinal discs, a slightly larger value than assumed in the 1961 model.

Because the erector spinae muscles act through a small moment arm, they must exert high forces to counteract flexion moments during lifting and/or forward bending.[10] These exertions induce high compressive forces on the spinal motion segments. (Note: A motion segment consists of two adjacent vertebral bodies, the intervertebral disc, and the connecting ligamentous structures. The motion segment has been studied extensively because it is the smallest segment of the spine that exhibits biomechanical characteristics of the entire spine.) Figure 1 summarizes compression forces acting at the L5/S1 spinal disc as a function of the load held in the hands and the horizontal reach distance. Under certain conditions of force and horizontal distance, compression forces may exceed limits recommended by NIOSH[33] and others.

Brinckmann et al.[52] argue that epidemiologic and clinical evidence show that some low-back problems are caused by primary mechanical destruction of tissues in the lumbar spine. Bone, cartilage, the intervertebral discs, and ligaments may fracture or rupture due to mechanical overload caused by excessive compression forces. Proof of this hypothesis is difficult to obtain, however, because it is unethical to perform in vivo experiments that expose the human body to mechanical loads at levels high enough to cause injury. Because of this dilemma, research has proceeded in two directions. The task-imposed load on the lumbar disc is either measured directly (see next paragraph) or calculated using biomechanical models. The load bearing capacity of spinal structures (bones, discs, and ligaments) must be determined in vitro using tissue obtained from cadavers. By comparing task-induced loads to load bearing capacity, it is possible to determine whether certain work activities exceed the mechanical limits of the spinal tissues.

Although most efforts to estimate spinal loads have been done using biomechanical models, there have been a few attempts to measure load in vivo. Nachemson and Elfstrom[4] developed a small pressure transducer that fits inside a hypodermic needle. The needle can be inserted into a spinal disc, allowing the transducer to measure the hydrostatic pressure inside the nucleus pulposis while a person assumes different postures and performs various tasks. In vitro experiments using this system with cadaver spinal segments showed a linear relationship between internal disc pressure and the external compressive load applied to the segment.[53] In vivo disk pressure measurements have been used successfully to validate biomechanical model predictions of compressive loads.[54] Due to the highly invasive nature of this technique, it is best suited for slow, well-controlled movements (similar to those considered by static biomechanical models) and can be used only under strict laboratory conditions. As a result, in vivo studies of intradiscal pressure have been few in number.

The tolerance of spinal motion segments to compressive loading has been studied extensively by biomechanical engineers and ergonomists in vitro using cadaver specimens. Genaidy et al.[35] reviewed 12 of the larger studies with the objective of combining the data sources to develop a statistical model of spinal compression tolerance limits. Their resulting equation ($R^2 = .83$) can be used to estimate lumbar compression strength limits for various percentiles of the population:

$$CS = 7222 - (1048 \times A) - (1279 \times G) + 57 \times PP$$

where CS is the load tolerance limit (Newtons); A is the age in decades (1 for 20–29 years, 2 for 30–39, 3 for 40–49, 4 for 50 years and above); G is gender (1 for males, 2 for females); and PP is population percentile (50th, 90th, 95th, etc.). Note that higher values represent higher tolerance.

Using this equation for a younger male with a strong lumbar spine (25 years old, 75th percentile), the resulting compression limit is 9170 N. For an older male with a weaker lumbar spine (55 years old, 25th percentile), the compression limit drops to 3176 N. For an older female with a weaker lumbar spine (55 years old, 25th percentile), the compression limit drops to 1897 N.

One of the limitations of cadaver studies is the uncertainty whether compression damage to spinal segments in vitro is a reliable predictor of the risk of injury associated with in vivo compressive forces during the performance of work tasks. In spite of this limitation, NIOSH has used epidemiological and biomechanical evidence to establish 3400 N as the maximum recommended L5/S1 disc compression force as one of the criteria used to develop the NIOSH lifting equation.[33,34] Citing a cadaver study,[55] NIOSH acknowledged the large variability in compression forces associated with disc failure. NIOSH estimated that 21% of spinal segment specimens would fail at 3400 N and that this limit may not protect the entire working population.[34]

The 3400 N criterion has been incorporated into several biomechanical models. Figure 1 presents L5/S1 compression forces predicted by The University of Michigan model for various combinations of weight and horizontal distance. Compression forces increase with weight and distance, and the NIOSH 3400 N limit can be exceeded when handling relatively light loads at an extended reach distance. (Note: Disc compression is the limiting criterion in the NIOSH lifting equation for infrequent lifting where fatigue is not a factor. For repetitive lifting, psychophysical and/or physiological factors become the limiting criterion due to fatigue.[33,34,48])

Most studies of spinal segment failure have utilized experimental procedures where the compressive load on the spinal segment was increased to the point of mechanical failure in a single trial.[35] A small number of studies have looked at the effects of repeated submaximal compressive loads on mechanical failure of the spinal segment.[55–57] In these studies loads were applied at a frequency of .25 Hz (15 cycles per minute). At compressive loads set at 55% of the single-trial failure load, mechanical failures were observed in 92% of the specimens after 5000 cycles. At a 65% load, failures occurred in 91% of the specimens after only 500 cycles. At a 75% load, some specimens failed after only 10 cycles. These numbers must be interpreted cautiously because cadaver tissue may respond differently to repeated loads than live tissue. However, these studies present limited evidence supporting frequency as a risk factor for some back injuries.

Summary of Static Biomechanical Models

Static kinematic models of the human musculoskeletal system have been used to evaluate strength capability and compressive forces acting on the lumbar spine during common whole-body manual materials handling tasks, such as lifting, pushing, and pulling. Large mechanical moments can be created in the lower back region by lifting heavy, compact loads close to the body or by lifting light-to-moderate loads at extended reach distances in front of the body. Forward bending also increases the mechanical moment due to the effects of body weight above the lower back. To counteract the moments created by loads in the hand and body weight, the extensor muscles of the lower back must exert high forces, creating a compression load on the lumbar spine. Based on biomechanical analysis, the critical task factors associated with lifting are (1) the amount of weight lifted, (2) the location of the load (horizontal distance from the lower back), and (3) body posture (forward bending of the trunk increases the load on the lower back).

The predictions of static biomechanical models are consistent with direct and indirect measurements of strain on musculoskeletal tissues. Electromyographic studies have shown that EMG activity in the erector spinae muscles increases with increased load in the hands and/or forward bent postures. Intradiscal pressure measurements have shown that hydrostatic pressure in the nucleus pulposis of the disc increases with increased load in the hands and/or forward bent postures. Cadaver studies of spinal motion segments have demonstrated mechanical failure of spinal tissues under compressive loads smaller than those predicted by biomechanical models for many lifting tasks. Furthermore, a small number of cadaver studies have demonstrated that cyclical loading reduces the mechanical tolerance limits of the lumbar spine, indicating that lifting frequency may be a factor in some back injuries. However, cadaver studies have been questioned because it is not known whether the behavior of cadaver tissue is similar to that of living tissue.

Although static whole-body biomechanical models are useful for evaluating stresses associated with common materials handling activities, they should not be used in all situations. The database of muscle strength capability in these models was collected using isometric tests of maximum strength capability. Because muscles cannot perform at maximum levels for extended periods of time or on a highly repetitive basis, the static biomechanical models tend to overestimate strength capability on jobs that require repeated exertions. For this reason, psychophysical and/or physiological-based methodologies are a preferred alternative for evaluating repetitive whole-body exertions.[58] Because static models do not consider forces and moments imposed on the musculoskeletal system from the acceleration/deceleration of external loads and body segment masses during highly dynamic movements, they may underestimate strain in work activities that involve rapid body motions.[59] Finally, the spine is an extremely complex mechanical structure. Without making many assumptions (such as using single equivalent muscles to represent multiple muscles) to simplify the system, solutions become statically indeterminate. As a result of these simplifications, the models are imprecise in evaluating certain stressors such as shear and torsional loads and the effects of coactivation of multiple muscle groups. Despite these limitations, static biomechanical models have provided good insight to some of the risk factors associated with manual handling activities.

Dynamic Whole Body Biomechanical Models

Dynamic biomechanical models for evaluating whole body materials handling activities have been developed by several investigators.[60–67] These models are inherently more complex than static models. In addition to considering external forces acting on the body (the load applied to the hands and effects of body weight) and posture, these models must also consider the effects of motion dynamics, including velocity and acceleration. Due to inertia, acceleration or deceleration of body segments and any load in hands requires the application of additional force as stated by Newton's second law (F = ma). Because the body is composed of multiple links and multiple joint centers, dynamic models require high-frequency measurements of many reference points to determine instantaneous locations, velocities and accelerations of model components. For this reason, dynamic models are often restricted to laboratory environments where accelerometers, goniometers, and/or motion analysis equipment can be used to collect reliable data. In spite of this limitation, dynamic models provide important insights into the additional biomechanical strain imposed by rapid motions.

Leskinen[65] used a simple two-link model (upper limbs plus trunk above the L5/S1) to compute compression forces at the L5/S1 spinal disc under static and dynamic assumptions. Input for the model was collected by obtaining posture and inertial data from 20 males who lifted a 15-kg box from a height of 10 cm to knuckle height. Model-predicted peak L5/S1 compression was 33 to 60% higher (depending on subject and lift technique) when the dynamic inertial load was added to the static model. Bush-Joseph et al.[66] used a dynamic model to compute peak moments at the L5/S1 spinal segment using data collected from 10 male subjects who lifted a 150-N box from floor level to a height of 1 m at slow, medium, and fast speeds. Peak moments increased linearly with the lift speed.

The effects of movement dynamics and other task factors on selected indices of biomechanical strain on the lower back were evaluated using a whole body model developed at the Universität Dortmund in Germany.[62] The basic representation of the skeletal system in the "Dortmunder" model is similar to the Michigan model described above; however, the lumbar spine is depicted as a system of five joints representing the five lumbar intervertebral discs (the Michigan model uses only one joint, the L5/S1, in the lower back region). The Dortmunder model was used to estimate the moment at the L5/S1 joint, compressive forces at the L5/S1 disc, and shear forces at the L5/S1 disc under various static and dynamic task conditions using 50th percentile male anthropometry during symmetric, sagittal plane lifting. The static analysis produced results consistent with the Michigan model. All three strain indices increased monotonically as the weight in the hands increased from a no-load condition to a 50-kg load. All three indices increased monotonically as the horizontal location of the load in front of the L5/S1 increased. With the hands empty, increased trunk flexion angle (forward bending) caused monotonic increases in L5/S1 moment, compression, and shear.

The dynamic analysis with the Dortmunder model demonstrated the significance of velocity and acceleration.[62] The task of raising objects of various weights (no load, 20, and 40 kg) from floor to elbow height was simulated in three conditions: "slow," in which the lift was completed in 2 sec; "medium," in which the lift was completed in 1.5 sec; and "fast," in which the lift was completed in 1 sec. In the slow condition, compressive forces at the L5/S1 were similar to static conditions. At medium speed, peak compressive forces were about 20% higher than under static conditions. At high speed, peak compressive forces were roughly 50% higher than under static conditions. Jager et al.[62] also simulated a jerking motion, assuming that all of the upward acceleration was completed in 0.1 sec. Under these conditions, the peak L5/S1 compression for lifting a 20-kg load from the floor was approximately 8000 N, more than double the peak load in static conditions and considerably higher than compressive loads shown to cause mechanical damage in cadaver tissues. Shear forces were not reported for the dynamic analyses.

Marras and Sommerich[63,64] of The Ohio State University developed at three-dimensional dynamic biomechanical model for evaluating loads on the lumbar spine during lifting. Instead of using a single equivalent trunk extensor muscle, this model extended earlier work[68] by including 10 functional muscle groups in the lower back. By considering multiple muscle groups, Marras and Sommerich were able to evaluate the effects of co-contraction and asymmetric postures on spinal stresses during lifting.

The Ohio State model was used to evaluate the effects of trunk velocity (10, 20, and 30 degrees/sec), trunk torque output (27.1 and 54.2 Nm), and trunk posture symmetry (symmetric, 30° clockwise rotation) on biomechanical loads on the spine. Eleven subjects participated in this test by performing isokinetic trunk extensions. During these exertions, electromyography measured muscle activity in 10 trunk muscles and was used as input to the model along with subject anthropometry and trunk kinetics. The model calculated compression, shear, and torsion loading in the lumbar spine. Muscle activity levels in the erector spinae were balanced between the left and right sides during symmetric lifting; however, during the asymmetric lift (trunk twisted clockwise), activity on the left side was dominant. In symmetric conditions peak compression at the L5/S1 increased with velocity (approximately 100 N for each increase of 10 deg/sec) and trunk torque output. Compression forces approached 3900 N at the 30 deg/sec, 54.3 Nm condition. In asymmetric conditions peak compression was level (approximately 3000 N) over the range of velocities tested and approximately 25% lower than under similar symmetric conditions. Peak anterior/posterior shear forces were greater under symmetric conditions and increased with the magnitude of trunk torque output. Left/right shear was approximately 40 N under asymmetric conditions compared with about 10 N under equivalent symmetric conditions.[63,64]

In a follow-up cross-sectional epidemiological study Marras et al.[59,68] used historical medical records of low-back injuries to classify over 400 cyclical jobs as either high risk or low risk. Dynamic trunk motions were measured for each job using a triaxial goniometer system (called the Lumbar Motion Monitor) to document three-dimensional angular position, velocity, and acceleration of the lumbar spine while workers performed their jobs. In addition, basic biomechanical variables (weight lifted, lift frequency, posture, etc.) were determined for each job. Logistic regression was used to identify the following five risk factors that discriminated between high- and low-risk jobs: (1) lifting frequency, (2) load moment, (3) trunk lateral velocity, (4) trunk twisting velocity, and (5) the trunk sagittal angle (odds ratio for five variables combined was 10.7).

The Ohio State group[69] used the Lumbar Motion Monitor in a laboratory study to evaluate the effects of lift symmetry and to compare one versus two-handed lift technique. Twenty-four subjects (all male) wore the Lumbar Motion Monitor while performing one and two-handed lifts ranging between 0° (in the sagittal plane) and 135°. Trunk motion characteristics associated with increased risk of back injury[59] were all higher with one-handed lifts, and velocities and accelerations increased substantially with the angle of asymmetry.

Kim and Chung[70] conducted a laboratory study of the electromyographic activity of the lower back during dynamic lifting. Eight healthy males participated in four 2-hour trials, lifting and lowering weights between floor and knuckle height. Independent variables were lift type (frequent-lifting, defined as lifting a weight normalized to 10% of the subject's strength six times per minute versus heavy-lifting, defined as lifting a weight normalized to 20% of the subject's strength three times per minute) and posture (symmetric versus 90° offset). Muscle activity (normalized EMG) was significantly higher during heavy lifting and during asymmetric lifting ($p < .001$). Muscle fatigue during the 2-hour trial (measured by a decrease in the mean power frequency of the EMG signal) was significantly greater during asymmetric and frequent lifting ($p < .001$).

Summary of Dynamic Biomechanical Models and Laboratory Experiments

Dynamic biomechanical models have been used by several investigators to overcome some of the limitations of the static models discussed previously. These models allow investigators to consider the effects of inertia and acceleration when estimating biomechanical stresses on the lower back during lifting activities. Under slow, controlled, sagittal-plane lifts (approximately 2 seconds from floor to elbow height), the Dortmunder[62] model shows that compressive forces acting at the L5/S1 spinal disk are similar to compressive forces under static conditions. By doubling the lifting speed, compressive forces increased by 50%. Studies performed at The Ohio State University show similar results for sagittal lifts; back compression increased with lifting speed. For asymmetric lifts, the Ohio model found unbalanced muscle activity between erector spinae muscles on the left and right sides, and increased lateral shear forces.[63,64]

Based on dynamic biomechanical analyses and limited epidemiological studies, the critical risk factors associated with lifting are (1) load moment about the spine (weight × horizontal distance for the handheld load and the weight of body segments above the L5/S1), (2) velocity of lift, (3) frequency of lift, (4) lift asymmetry (lateral and twisting velocities), and (5) the trunk sagittal flexion angle. At faster lifting speeds relatively light loads in the hand can result in back compression forces that exceed the 3400-N threshold level established by NIOSH.

Dynamic models are not appropriate for all lifting situations. For highly repetitive lifting, dynamic models do not consider the effects of fatigue; therefore, psychophysical and/or physiological methods are preferred. Due to the increased complexity of the data collection and analysis when using dynamic models, static approaches may prove to be more practical if lifts are performed using slow, controlled motions.

PSYCHOPHYSICAL STUDIES OF MANUAL MATERIALS HANDLING

Psychophysical studies of lifting and related manual materials handling activities have focused on workers' perceptions of physical strain, discomfort, and fatigue associated with work. There are several distinct differences between the biomechanical methods previously discussed and the psychophysical methods discussed in this section. First, psychophysical methods are used to measure subjective responses to work (discomfort, fatigue, etc.), whereas biomechanical methods focus on objective responses (EMG activity, disc compression, etc.) Second, biomechanical approaches are primarily concerned with predicting how body tissues react during a single exertion, whereas psychophysical methods can be used to assess how workers respond to work demands that are distributed over a shift of 8 or more hours.

The Liberty Mutual Studies

The Liberty Mutual experiments (two-handed, repetitive tasks performed for an extended work period)[22] were all performed in a laboratory setting using subjects recruited from local industries (evening shift workers) near Hopkinton, Mass. Subjects performed a variety of common materials handling tasks, such as lifting, carrying, pushing, and pulling. They were instructed to perform as if they were being paid on an incentive basis, working as hard as they could without becoming unusually tired, weakened, overheated, or out of breath. Instead of measuring maximum strength in a single exertion, this approach measured exertions that could be performed on a repeated basis over an extended period without excessive fatigue or discomfort. Subjects were given control over one task variable, the weight (or resistance force for pushing and pulling tasks) of the object being handled. They could increase or decrease this weight at will. All other variables, such as task frequency, initial height of load, distance lifted/lowered/carried, etc. were controlled by the experimenter. For each condition tested, the subjects adjusted the weight to the maximum amount they would be willing to handle if the task were performed throughout an 8-hour work shift. (Note: The duration of the test session was typically 4 rather than 8 hours.) The final weight (following adjustments) was recorded as the MAW. All tasks were performed with two hands symmetric to the sagittal plane. Liberty Mutual has published extensive tables of MAWs for various task conditions.[22] Separate tables exist for males and females, listing maximum acceptable weights for various percentiles of the working population.

The Liberty Mutual studies can be summarized by the following points:

■ *Two-handed symmetric lifts.* MAW decreases as the following task variables increase: lift frequency, lift distance (displacement), and object size. In addition, MAWs are lower when lifting is above shoulder height than when lifting is in the floor-to-knuckle-height or knuckle-to-shoulder height ranges.

■ *Two-handed symmetric lowers.* MAW decreases as the following task variables increase: lift frequency, lift distance (displacement), object size. In addition, MAWs are lower when lifting is above shoulder height than when lifting is in the floor-to-knuckle-height or knuckle-to-shoulder height ranges.

■ *Two-handed pushes.* Maximum acceptable force (MAF) decreases as the following task variables increase: push frequency and push distance. Push forces are highest when the handles are located at approximately elbow height. MAF is reduced when the handles are located above shoulder level or below knuckle height.

■ *Two-handed pulls.* MAF decreases as the following task variables increase: pull frequency and pull distance. Pull forces are highest when the handles are located at approximately knuckle height. MAF is reduced when the handles are raised to elbow height and reduced again when the handles are raised above the shoulder.

■ *Carrying.* MAW decreases as task frequency and carry distance increase. MAW is greater when carrying at waist level compared with carrying at midchest level.

The Liberty Mutual tables show a large variance in population capability. Within gender, MAWs and MAFs demonstrated by the strongest 10% of the population were roughly double those demonstrated by the weakest 10%. Males as a group were stronger than females, but there was considerable overlap between the genders.

Relationship Between Liberty Mutual Results and Workplace Injuries

Snook et al.[71] performed a retrospective study of 191 compensable work-related low-back injuries from 32 states. Physical demands on the jobs where injuries occurred were evaluated and compared with the Liberty Mutual psychophysical database to estimate the percentage of workers who would find the demands acceptable. Workers assigned to jobs in which physical demands exceeded the level deemed acceptable by 75% of the population were three times as likely to experience a back injury when compared with workers on jobs in which physical demands were below the level acceptable to 75% of the population. Based on the distribution of workers in the 75%-and-above versus below 75% jobs, the authors concluded that up to one-third of compensable back injuries could be prevented by designing jobs to fit at least 75% of the population.

The Liberty Mutual psychophysical tables were used to evaluate the physical demands on the jobs of 6912 workers in 5 different industries.[47] A significant negative correlation ($p < .05$) was found between overexertion injury incidence rates and psychophysical "percentage capable" rating for the most stressful task on these jobs (i.e., as percentage capable increased, injury rates decreased). Similar to the study described in the previous paragraph,[71] there was a background level of overexertion injuries that could not be attributed to physical demands.

These studies[47,71] provide evidence supporting the utility of using psychophysical percentage capable scores for identifying manual materials handling activities that place workers at increased risk of overexertion injury. NIOSH included psychophysical criteria (along with biomechanical and physiological criteria) in the development a quantitative tool for evaluating stresses associated with manual lifting. The original[33] and revised[34,48] NIOSH lifting equations establish weight limits that are acceptable to 75% of adult females and 99% of adult males.

Studies of Two-Handed Tasks During 12-Hour Work Shifts

A University of Cincinnati study [19, 20] replicated a subset of the Liberty Mutual experiments to investigate the effects of an extended work shift (12 hours) on MAW and energy expenditure. Independent variables were lift frequency, height of lift (floor-to-knuckle, knuckle-to-shoulder, and shoulder-to-reach), box size, and shift length. For a given condition of frequency, height, and box size, subjects selected MAWs first assuming an 8-hour shift and then assuming a 12-hour shift. Similar to the Liberty Mutual studies, the following results were significant ($p < .05$): (1) MAW decreased as lift frequency increased; (2) MAW decreased as box size increased; (3) MAW was greater at the floor-to-knuckle height than at the shoulder-to-reach height; and (4) MAW was greater for males than females.

Over all conditions tested, the MAW for males decreased by an average 22% when going from an 8 to 12-hour shift ($p < .05$). Although the MAW decreased by an average of 12% for females, the difference was not statistically significant. Energy expenditure rates, based on measurement of oxygen uptake and computed as a percentage of aerobic capacity, decreased with the longer work shift. For males, average energy expenditure decreased from 29 to 23% of aerobic capacity. For females, the corresponding change was from 28 to 24% of aerobic capacity. Neither change was statistically significant.

Studies of Asymmetric Lifting

In studies at The University of Wisconsin[72] the psychophysical method was used to compare MAW for two-handed floor-to-table (81 cm) lifts in the sagittal plane versus two-handed floor-to-table lifts at asymmetry angles of 30, 60, and 90° in a laboratory study of 18 male college students. MAW decreased by 7, 15, and 22% as the asymmetry angle deviation from the sagittal plane increased ($p < .05$). MAW also decreased with larger box sizes ($p < .01$). In a follow-up experiment the effects of lift symmetry, frequency, and height of lift on MAW were evaluated. MAW at angles of 30, 60, and 90° was compared with MAW in the sagittal plane with mean observed decreases of 9, 14, and 21% ($p < .01$).[73] In addition, MAW decreased with increasing lift frequency ($p < .01$).

At The University of Cincinnati,[21] the psychophysical method was used to determine the effects of lift symmetry, load symmetry, load size, frequency, and height of lift on MAW. Heart rate and oxygen uptake also were monitored during the experiment. Major findings included the following:

■ Asymmetric lifting (turning 90° from the sagittal plane) reduced MAW by an average 8.4% ($p < .01$).

■ Asymmetric loading (offsetting the center of gravity of the load 20 cm from the sagittal plane) reduced MAW by an average 10% ($p < .01$).

■ Similar to the Liberty Mutual studies, MAW decreased significantly ($p < .01$) with increased lift frequency and increased object size. MAW was lower when lifting above the shoulder compared with lifting floor-to-waist. There was also a significant subject effect ($p < .01$) reflecting the range of capabilities in the subject pool.

Neither lift symmetry nor load symmetry had a significant effect on heart rate or oxygen consumption.

Studies of the Effects of Handles and Container Shape

Garg and Saxena[74] used the psychophysical method to determine the effects of handles, container shape, and container dimensions on MAW in a laboratory study of 10 college students at the University of Wisconsin. Subjects lifted six different-sized tote boxes and three different sized mail bags from the floor to a 76-cm bench. Each of the tote boxes had two configurations—with and without handles. The lack of handles on boxes decreased MAW by 7.2% averaged across all box sizes and subjects ($p < .01$). MAW for the smallest box (38×51 cm) was on average 10% greater than MAW for the largest box (64×64 cm) for both handle and no-handle conditions ($p < .01$). MAW averaged across all mailbags was greater than the average MAW for no-handle boxes but less than the average MAW for handle boxes ($p < .01$).

The effects of handles versus no handles on MAW also has been studied at Liberty Mutual. When lifting boxes without handles, MAW was consistently lower (median reduction of 16%) compared with lifting similar boxes equipped with handles.[75]

Studies of MAWs in Restricted Work Postures

The psychophysical method was used by the U.S. Bureau of Mines to determine MAW under conditions of restricted headroom where it is impossible for a worker to stand fully erect, such as when working in low-ceilinged coal mines.[76] Eight experienced coal miners served as subjects in a laboratory study to evaluate the effects of posture (kneel versus stoop), lift symmetry, and lift distance (35 cm versus 60 cm) on MAW. All tests were performed under a 1.22-m ceiling to prevent the subjects from standing. Mean MAW was reduced by 11% in the kneeling posture compared with stooping ($p < .05$). MAW was significantly greater under asymmetric conditions ($p < .01$) and at the smaller lift distance ($p < .05$); however, the relative differences were small (less than 5%).

Smith, et al.[23] used the psychophysical method to determine MAW when lifting, lowering, or carrying in nonstandard postures such as twisting, lying down, kneeling, squatting, and work with restricted ceiling heights. One hundred subjects (50 men, 50 women) recruited from a college-age population participated in this laboratory study at Texas Tech University. Although this study did not include measurements of MAW in normal lifting (two-handed, symmetric, sagittal-plane) to use as a basis of comparison, the following trends were observed:

- MAW decreased with increased lift height.
- MAW decreased with twisting.
- MAW decreased in one-handed lifts (compared with similar two-handed lifts).
- MAW decreased during carrying tasks with lower ceiling heights.

Comparison of Psychophysical Findings with Other Criteria for Job Design

Several studies have been performed to compare MAWs established using the psychophysical approach against energy expenditure and biomechanical criteria for designing manual materials handling tasks. Investigators have compared energy expenditure when working at psychophysically determined MAWs to the NIOSH[33] recommendation of 3.5 kcal/min to avoid excessive physiologic fatigue. These studies[77–79] have shown that at rapid lifting frequencies (4.3 lifts per minute or faster), psychophysically determined MAWs exceed the NIOSH-recommended energy expenditure levels. For low-to-moderate lifting frequencies, energy expenditure when working at the MAW was below the 3.5 kcal/min criterion. At the other extreme, MAWs determined for very low lifting frequencies (once every 5 min) have been used to compute spinal compression forces using biomechanical models. A recent analysis[80] of floor-to-knuckle-height lifts found that the once-per-5-min lift MAW acceptable to a relatively large proportion of the working population (90% of adult males) resulted in L5/S1 compression forces that exceed the 3400 N limit recommended by NIOSH.[34]

Questions also also been raised regarding whether MAWs established during a relatively short experimental session (typically on the order of 30 min) are valid for extended work periods. Mital,[81] for example, found decreases in MAW at the end of an 8-hour experimental session when lifting at frequencies greater than six lifts/minute. Ciriello et al.[80] found stability in MAWs during 4-hour sessions as long as the lifting frequency was slower than 4.3 lifts/min.

Summary of Psychophysical Studies of Manual Materials Handling

The psychophysical method has been used in a series of laboratory studies to determine MAW (for lifting, lowering, and carrying tasks) and MAFs (for pushing and pulling tasks). Using this approach, subjects adjust the level of weight they are willing to lift (or the level of force they are willing to exert) as task requirements change. These studies have demonstrated that the following task factors are significant:

- *Frequency (repetition)*. As task frequency increases, MAW and MAF are reduced.
- *Posture (vertical location)*. MAW and MAF are reduced for tasks performed above shoulder height.
- *Posture (asymmetry)*. MAW is reduced when lifting activities are performed outside the sagittal plane, requiring twisting of the trunk.
- *Displacement (travel distance)*. MAW and MAF decrease as the travel distance of the handled object increases.
- *Object size*. MAW is reduced as the size of the lifted object increases.
- *Handles*. MAW decreases when handles are not provided.
- *Shift length*. MAW decreases when moving from an 8 to 12 hour shift.
- *Individual capability (subject effects)*. Within the working population, there is a wide range in individual capability to perform manual materials handling tasks. Although gender is statistically significant (MAW and MAF are greater for males), there is substantial overlap between the male and female distributions.

There is limited evidence[47,71] supporting the use of psychophysical guidelines for designing manual materials handling tasks. Potentially up to one-third of compensable back injuries could be prevented by designing jobs to accommodate at least 75% of the working population.

SUMMARY OF BIOMECHANICAL AND PSYCHOPHYSICAL STUDIES OF RISK FACTORS RELATED TO LOW-BACK PAIN

This article has reviewed recent biomechanical and psychophysical research on workplace factors associated with low-back pain. The biomechanical studies have examined the relationship between selected work parameters (e.g., weight lifted, reach distance, posture) and selected strain responses of body tissue (e.g., electromyographic activity of muscles, intradiscal pressure, job

TABLE II. Summary of Job and Task Factors Significantly Related to Biomechanical and/or Psychophysical Measures of Strain

Job or Task Factor	Biomechanical Strain	Psychophysical Strain
Weight lifted	X	(Dependent Variable)
Horizontal reach distance	X	X (Box Size)
Posture: trunk flexion	X	
Posture: trunk twisting/bending	X	X
Posture: lift above shoulder		X
Lift frequency	X	X
Lift dynamics (acceleration)	X	
Displacement distance		X
Presence of handles		X
Shift duration		X
Population variability	X	X

strength requirements versus worker strength capabilities). The psychophysical studies have examined the relationship between selected work parameters and the amount of weight that people are willing to handle without excessive fatigue.

Table II presents a summary of task characteristics found to be significantly relevant to one or more biomechanical or psychophysical measures. Considerable additional research is needed to understand the quantitative relationships between exposure to these factors and the incidence and severity of work-related overexertion injuries and disorders of the lower back. Nonetheless, these task attributes should be considered during job evaluation and job design procedures in order to reduce exposures to those factors proven to cause increased biomechanical and/or psychophysical strain.

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