Introduction

Low back disorders (LBDs) are highly prevalent worldwide, affecting up to 85 per cent of adults at some time in their lives, causing suffering, disability and loss of productivity (Frymoyer, 1996). The one-year prevalence rate in the US, Germany, Norway, and Sweden, for instance, has been reported to be as high as 56 per cent, 59 per cent, 61 per cent and 70 per cent, respectively (Ihlebaek et al., 2006; Manchikanti, 2000; Schneider et al., 2007). They remain as a major economic burden on individuals, industries and societies as a whole. In the US during 2005 alone, the total cost associated with LBDs has been suggested to vary from $100 to $200 billion (Katz, 2006) that is comparable with an estimated $81.2 billion in damage associated with the Hurricane Katrina, recognized as the costliest natural disaster in the US history, that took place in the same year (Wikipedia Encyclopedia, 2006).

Although low back pain (LBP) could originate from different musculoskeletal structures such as vertebrae, ligaments, facet joints, musculature, and disc annulus fibrosis, in most cases the exact cause of the symptoms remains, however, unknown (Diamond and Borenstein, 2006). In a large survey, lifting or bending episodes accounted for 33 per cent of all work-related causes of LBP (Damkot et al., 1984). Combination of lifting with lateral bending or twisting that occurs in asymmetric lifts has been identified as a frequent cause of back injury in the workplace (Andersson, 1981; Hoogendoorn et al., 2000; Kelsey et al., 1984; Marras et al., 1995; Troup et al., 1981; Varma and Porter, 1995). Among various work-related activities, lifting, awkward posture, and heavy physical work have strong relationship with lumbar musculoskeletal disorders (NIOSH, 1997). Lifting, therefore, is one of the major documented risk factors for LBDs (Burdorf and Sorock, 1997; Ferguson and Marras, 1997; Frank et al., 1996).

The foregoing studies confirm an association between manual material handling tasks and LBDs and suggest that excessive and repetitive mechanical loads acting on the spine could play major causative roles in LBDs. Proper knowledge of ligamentous loads, muscle forces, and trunk stability in the normal and pathologic human spine under various recreational and occupational activities, hence, becomes crucial towards appropriate and effective
management of LBDs. Prevention, rehabilitation, treatment, and performance enhancement programs stand to substantially benefit from an improved understanding on the load partitioning in the human spine. Infeasibility of direct measurement of muscle forces and spinal loads in human beings and the limitations in extrapolation of such data collected from animal studies have led to indirect quantification of loads on spine by measuring representative biomedical indicators (e.g. intra-discal pressure, muscle electromyographic (EMG) activity). However, apart from invasiveness, cost concerns, limitations and difficulties, the validity of such indicators to adequately represent spinal loads has also been questioned (van Dieen et al., 1999). Biomechanical models have, thus, been recognized as indispensable tools for estimation of muscle forces, spinal loads, and trunk stability during various occupational and recreational activities.

**Single-level biomechanical models**

Biomechanical models, both static and dynamic, use basic principles of mechanics to estimate muscle forces and spinal loads under different loading conditions. Forces in various active (i.e., trunk muscles) and passive (e.g. posterior ligaments and discs) structures are calculated by consideration of equilibrium equations. A free body diagram of the trunk (typically cut by an imaginary plane through the L4-L5 disc) is employed to maintain equilibrium between known external moments (due to gravity/inertia/external loads usually estimated by a link segment model) and unknown internal moments (due to spinal active and passive structures at the plane of cut). Unfortunately, such equilibrium equations cannot be resolved deterministically, as the number of unknowns significantly exceeds that of available equations (*kinetic redundancy problem*). A number of biomechanical models have been introduced to tackle the foregoing kinetic redundancy in equilibrium equations and to estimate spinal and muscle loads. Three approaches that have often been used in the analysis of different joint systems are: single-equivalent muscle, optimization-based, and EMG-assisted approach (Gagnon et al., 2001; Shirazi-Adl and Parnianpour, 2001; van Dieen and Kingma, 2005).

In the reduction or equivalent muscle approach, the role of muscles is simplified by neglecting some and grouping others into synergistic ones, assuming a priori known activation levels; thus reducing the number of unknown muscle forces to the available equilibrium equations. In the optimization approach it is assumed that there is one cost (objective) function (or many cost functions) that may be minimized or maximized by the central nervous system (CNS) while attempting to satisfying the equilibrium conditions. Constraint equations on muscle forces are introduced in parallel enforcing that muscle forces remain greater than zero and smaller than some maximum values corresponding to the maximum allowable stress in muscles. Various linear (e.g., related to axial compression) and nonlinear (e.g., related to muscle fatigue) cost functions have been employed. The nonlinear cost function of the sum of cubed or squared muscle stresses has been suggested to adequately match collected EMG data (Arjmand and Shirazi-Adl, 2006a). In the EMG-assisted approach, electromyography signals of limited and often only superficial trunk muscles are first measured. A relationship between normalized EMG activity of a trunk muscle and its force is subsequently presumed, allowing for the estimation of individual trunk muscle forces while satisfying the existing equilibrium equations (Gagnon et al., 2001; Marras and Granata, 1997; McGill and Norman, 1986). Each of these three approaches has its own advantages and drawbacks (van Dieen and Kingma, 2005; Reeves and Cholewicki, 2003).
Multi-level biomechanical models

Single-level models have been and remain to be very popular in biomechanical model investigations of different multi-joint studies (e.g., Cholewicki et al., 1995; Granata et al., 2005; Marras et al., 2006; McGill and Norman, 1986; Parnianpour et al., 1997; Schultz et al., 1982; van Dieen et al., 2003; van Dieen and Kingma, 2005). These models have widely been employed in ergonomic applications and in injury prevention and treatment programs. A major shortcoming with these models, however, lies in the consideration of the balance of net external moments only at a single joint or cross-section (typically at lowermost lumbar discs) rather than along the entire length of the spine. This drawback naturally exists in dynamic and quasi-static model studies alike while simulating either sagittally symmetric two-dimensional (2D) or asymmetric three-dimensional (3D) movements. It has been demonstrated that the muscle forces evaluated based on such single-level equilibrium models, once applied on the system along with external loads, will not satisfy equilibrium at remaining levels along the spine (Arjmand et al., 2007). They will neither yield the same deformed configuration based on which they were initially evaluated.

To overcome the foregoing major shortcoming, multi-level stiffness model studies, along with optimization (Gardner-Morse et al., 1995) or EMG-assisted (Cholewicki and McGill, 1996) approaches, have been developed and used to evaluate muscle recruitment, internal loads, and stability margin. The former model neglects nonlinearities in spinal behavior whereas the latter overlooks translational DoF at various joint levels and, hence, associated shear/axial equilibrium equations. These omissions have been found to adversely influence predictions on muscle forces and spinal loads (Arjmand, 2006).

Kinematics-driven approach

For more than a decade, our group has been developing a novel iterative Kinematics-driven approach in which a-priori measured vertebral/pelvis rotations of the spine (i.e., movement trajectory as much as available) are prescribed a priori into a nonlinear finite element model to evaluate muscle forces, internal loads, and spinal stability in static and dynamic analyses of lifting tasks with and without loads in hands (Arjmand and Shirazi-Adl, 2006b; Bazrgari et al., 2007; El-Rich et al., 2004; Kiefer et al., 1997; Shirazi-Adl et al., 2002). This iterative approach (see Figure 3.1) not only satisfies the equilibrium equations in all directions along the entire length of the spine but yields spinal postures in full accordance with external/inertia/gravity loads, muscle forces, and passive ligamentous spine with nonlinear properties. Using this approach, the role of the intra-abdominal pressure (IAP) (Arjmand and Shirazi-Adl, 2006c), lumbar posture (Arjmand and Shirazi-Adl, 2005) and lifting technique (Bazrgari et al., 2007; Bazrgari and Shirazi-Adl, 2007) on loading and stabilization of the spine during lifting activities has also been examined. Moreover, wrapping of thoracic extensor muscles around vertebrae while taking into account the contact forces between muscles and remaining spinal tissues in between has been simulated for the first time by using the Kinematics-driven approach (Arjmand et al., 2006).

In this approach, a sagittally-symmetric nonlinear finite element model of the entire thoraco-lumbar spine has been considered. This is a beam-rigid body model comprising of six deformable beams to represent T12-S1 discs and seven rigid elements to represent T1-T12 (as a single body) and lumbosacral vertebrae (L1 to S1) (Figure 3.2). The beams model the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments)
at different directions and levels. The nonlinear load-displacement response under single and combined axial/shear forces and sagittal/lateral/axial moments, along with the flexion versus extension differences, are represented in this model based on numerical and measured results of previous single- and multi-motion segment studies (Arjmand and Shirazi-Adl, 2005, 2006b). The trunk mass and mass moments of inertia are assigned at gravity centers at different levels along the spine based on published data for trunk segments and head/arms. Connector elements parallel to deformable beams are added to account for the intersegmental damping using measured values; translational damping $= 1,200 \text{ N s/m}$ and angular damping $= 1.2 \text{ N m s/rad}$ (Bazrgari et al., 2007).

Figure 3.1 Flow-chart for the application of the Kinematics-based approach to predict trunk muscle forces, spinal loads, and stability (critical lever arms for thoracic muscles can be defined as a function of their initial values in upright posture; anatomy of muscles used in the model is depicted in Figure 3.2, convergence is attained if calculated muscle forces in two successive iterations remain almost the same).
In the present study, the Kinematics-driven approach is applied to analyze the steady state flexion relaxation phenomenon that remains as a controversial issue in biomechanics of the spine in forward flexion postures.

**Application: flexion relaxation phenomenon (FRP)**

Upon progressive forward flexion of the trunk from the upright standing posture towards the peak flexion, a partial or complete silence in EMG activity of superficial extensor muscles has been recorded. This phenomenon has been well documented in healthy asymptomatic subjects and is called the flexion–relaxation phenomenon (FRP) (Floyd and Silver, 1951) that may persist even in the presence of weights carried in hands. The FRP has been recorded to occur at about 84–86 per cent of peak voluntary flexion in slow movements, irrespective of the magnitude of load carried in hands (Sarti *et al*., 2001). The presence and absence of the FRP could be used as a signature to discriminate LBP patients from healthy controls, as in the former group the FRP is frequently absent (Kaigle *et al*., 1998; Kippers and Parker, 1984; Watson *et al*., 1997). The FRP assessment has, thus, been suggested as a valuable clinical tool to aid in the diagnosis and treatment of LBP patients (Colloca and Hinrichs, 2005).
In order to explain the partial or full relaxation in back muscles in large trunk flexion postures, several hypotheses have been put forward in which the load is transferred from extensor muscles to passive tissues (Floyd and Silver, 1951; McGill and Kipper, 1994), from superficial muscles to deeper ones (Andersson et al., 1996), or from lumbar extensors to thoracic ones (Toussaint et al., 1995). Since the FRP is likely related to the relatively large axial strain (or elongation) in extensor muscles during forward flexion, it is expected to also depend on the lumbar rotation and pelvic-lumbar rhythm. The relative activity of various back muscles in deep flexion movements remain controversial as some suggest relaxation in global extensor muscles (Mathieu and Fortin, 2000; Sarti et al., 2001) while others report relaxation only in lumbar extensor muscles (McGill and Kipper, 1994; Toussaint et al., 1995). Using deep wire electrodes, Andersson et al. (1996) reported silence only in superficial lumbar erector spinae muscles with activity remaining in deeper ones.

Method

The trunk and pelvic rotations required in the Kinematics-driven approach to analyze the trunk movement in deep flexion was obtained from our own parallel ongoing in vivo measurements performed on 14 healthy male subjects with no recent back complications. For this purpose, infrared light emitting markers, along with a three-camera Optotrak system (NDI International, Waterloo/Canada), were used to track movement trajectory at different joint levels. From upright standing postures, subjects were instructed to slowly flex the trunk forward as much as possible. Total trial duration lasted about 12 seconds, including three seconds of rest at upright and peak flexion positions and six seconds of slow forward movement in between. Measured trunk and pelvic rotation trajectories for a typical subject were prescribed into the transient finite element model at the T12 and S1 levels, respectively. As for the individual lumbar vertebrae, the total lumbar rotation evaluated as the difference between foregoing two rotations was partitioned between different levels in accordance with proportions reported in earlier investigations (Arjmand and Shirazi-Adl, 2006b). Kinematics-driven approach was subsequently employed to calculate muscle forces, spinal loads and system stability margin throughout forward flexion movement (Figure 3.1).

In the current study, the cost function of the minimum sum of the cubed muscle stresses was considered in the optimization with inequality equations of muscle forces remaining positive and greater than their passive force components (calculated based on muscle strain and a tension-length relationship (Davis et al., 2003)) but smaller than the sum of maximum physiological active forces (taken as 0.6 MPa × PCSA where PCSA is the physiological cross-sectional area) and the passive force components. The finite element program ABAQUS (Hibbit, Karlsson and Sorensen, Inc., Pawtucke, RI, version 6.5) was used to carry out nonlinear transient structural analyses while the optimization procedure was analytically solved using an inhouse program based on Lagrange multipliers method (LMM). The total computed muscle force in each muscle was partitioned into active and passive components with the latter force evaluated based on a length-tension relationship (Davis et al., 2003).

Parametric studies on passive properties

The choice of a passive muscle length-force relationship used in the model for all muscles would influence the magnitude of muscle activity and, hence, the appearance or absence of
partial or full flexion relaxation. Due to the existence of a number of rather distinct curves in the literature on the muscle passive length-tension relationship, it was decided to alter the reference curve used in the model (Davis et al., 2003) and compute its effect on results. In this work, the passive curve was shifted either by +5 per cent (softened curve) or by –5 per cent (stiffened curve) in muscle axial strain and the effects of such decrease or increase in passive muscle resistance on results were analyzed. Furthermore, to investigate the relative importance of ligamentous spine passive properties on predictions, the bending stiffness of motion segments was also altered by ±20 per cent and the analyses were repeated. A decrease in bending stiffness (–20 per cent) could approximately simulate segmental degeneration, injury or tissue viscoelasticity, whereas an increase (+20 per cent) could simulate stiffer segments or ones with constructs or bone fusion.

Results

The trunk, pelvis, and lumbar rotations used in the model reached their peak values, though not at the same time, of 127.4°, 76.4°, and 51.9°, respectively (see Figure 3.3 for temporal variations of these prescribed rotations). Starting from the upright posture, the initial trunk rotation is due primarily to the lumbar rotation that reaches its maximum early in flexion phase of the movement. Subsequently, further increase in the trunk rotation at the mid- and final phases of flexion movement is found to be due solely to the pelvic rotation.

Starting from the upright standing posture towards full trunk flexion, temporal variation of active force in abdominal and global extensor muscles are also shown in Figure 3.3. The global extensor muscles experience an initial increase in activity as the forward flexion

Figure 3.3 Predicted temporal variation of active and passive forces in global trunk muscles (on each side) as well as measured trunk and pelvic rotations with time advancing from upright standing posture (left) toward full flexion (right). Total lumbar rotation is the difference between trunk and pelvic rotations. The initial drop in global extensor muscle forces is due to the inertia of the upper body at the onset of the task.
initiates followed by a decrease and finally a complete silence at larger flexion angles. In contrast, the abdominal muscles are active only at larger flexion angles when the thoracic extensor muscles are silent. The passive contribution of thoracic extensor muscles monotonically increases with flexion reaching their maximum at full lumbar flexion (Figure 3.3). Local lumbar muscles, with the exception of the multifidus, demonstrate a pattern similar to that of global thoracic muscles but with the full relaxation at larger trunk angles. The multifidus, on the other hand, undergoes only a partial relaxation (see Figure 3.4 for the L3 level).

The ligamentous spinal loads in the local segmental directions (i.e., axial compression, anterior-posterior shear force, and bending moment), as well as net external moment at the L5-S1 disc level, are also shown in Figure 3.5 for the entire duration of movement. The spinal loads increase downward reaching their maximum values at the distal L5-S1 level. The net external moment increases with the trunk flexion except at larger values where it drops slightly. The local axial compression and shear forces also follow the same trend, reaching peak values of 2629 N and 689 N, respectively.

As the passive contribution of muscles increases (i.e., case with −5 per cent shift in passive force-length relationship shown in Figure 3.6), the activity in thoracic extensor muscles substantially diminishes, demonstrating the FRP at smaller trunk flexion angles while activity in abdominal muscles initiates earlier at smaller trunk flexion and increases further at larger flexion angles. In contrast, activity in thoracic muscles increases as the passive contribution decreases, resulting in a delay in the FRP and a residual activity in the longissimus muscle at peak flexion angles. Similar trends, but to a lesser extent, in activities of thoracic extensor muscles and abdominal muscles are predicted when the passive properties of the ligamentous spine are increased or decreased by 20 per cent, respectively (Figure 3.7).
Discussion

Excessive mechanical loading of the spine is recognized as a major cause of LBDs. To reduce the risk of back injuries, existing ergonomic guidelines aim to limit compressive force on the spine (Waters et al., 1993). In the absence of any direct method to measure spinal loads and muscle forces, biomechanical modeling of the spine has become an indispensable tool in evaluation, prevention, and proper management of back disorders. A number of mathematical biomechanical models of the spine with diverse simplifying assumptions have been

Figure 3.5 Predicted temporal variation of passive ligamentous loads as well as net external moment at the L5-S1 disc level with time during forward flexion movement.

Figure 3.6 Predicted temporal variation of active force (on each side) in extensor global muscles with time during forward flexion movement as passive force-length curve is altered ±5% for all muscles.
introduced to address related biomechanical and clinical issues. Our earlier studies have demonstrated the crucial role of consideration of equilibrium at all levels and not just one, both translational and rotational degrees of freedom at each joint, wrapping of global extensor muscles, and nonlinearity of passive ligamentous spine on predictions.

The novel Kinematics-driven model takes into account the nonlinear behavior of the ligamentous spine while satisfying equilibrium conditions at all thoracolumbar spinal levels and directions. This approach has previously been applied to address a number of biomechanical issues during static and dynamics lifting activities in both upright and forward flexed standing postures. In the current study, though we used a full transient analysis, due to the rather slow pace of the forward flexion movement, a quasi-static approach would have yielded almost the same results. Consideration of inertia is particularly essential when the task is performed at movement velocities much faster than that considered in this study.

Physical activities involving trunk flexion are very common in regular daily, occupational, and athletic activities. An improved understanding of spinal functional biomechanics during full trunk flexion (i.e., partitioning of loads between passive spine and active trunk muscles and between active-passive components of muscles) are essential in proper analysis of risks involved. The kinematics required as input data into the finite element model have been taken from an ongoing in vivo study on the effect of movement velocity on trunk biomechanics. Throughout a forward flexion movement from upright standing posture to full flexion, muscle forces (active and passive), ligamentous spinal loads (axial compression force, anterior-posterior shear force, and sagittal moment) at all levels have been estimated. The effect of alterations in muscle passive force-length relationship and in bending properties of the ligamentous spine on results is also investigated.

During forward flexion movement simulated in this study, a sequential lumbar-pelvic rotation is observed in which greater lumbar rotation is apparent at the beginning of the task followed by pelvic rotation at the final phase (Figure 3.3). Similarly, as compared to the lumbar rotation, the pelvic rotation has been reported to become predominant at the end of flexion and beginning of extension phase during flexion-extension movements.
(Paquet et al., 1994). Others, however, suggest that lumbar and pelvic rotations act simultaneously during flexion and/or extension phases (Nelson et al., 1995; Sarti et al., 2001). The lumbar-pelvic rhythm affects results by influencing the wrapping of global extensor muscles and the relative passive contribution of muscles and ligamentous spine.

As the trunk flexes forward from upright posture, initially both active and passive components of forces in global extensor muscles increase with the formers reaching their peak values at about 45° (Figure 3.3). Thereafter, up to the trunk flexion of about 95°, active forces in thoracic extensor muscles diminish despite the continuous increase in net external moment reaching its maximum of 118 Nm. On the contrary, passive muscle forces as well as passive ligamentous moment increase throughout the movement to peak lumbar flexion (Figures 3.3 and 3.5). The progressive relief in activity of global back muscles is due, therefore, to higher passive contribution of muscles and ligamentous spine as the lumbar rotation increases. As the trunk flexion exceeds about 95° (at about 3.3 sec), lumbar rotation (Figure 3.3) and consequently both passive muscle force and moment resistance of the ligamentous spine, remain nearly unchanged, while the activity of back muscles continues to drop. In this case, the reduction in net external moment due to the decrease in the effective lever arm of the trunk centre of mass (COM) is the primary cause in progressive decrease in back muscle activities. Global longissimus [LGPT] and iliocostalis [ICPT] become completely silent at trunk flexion angles of about 114° and 95°, respectively.

With the exception of the multifidus that only partially relaxed, local lumbar muscles also demonstrated full relaxation in activity, but at larger flexion angles as compared with global extensor muscles (Figure 3.4). Measurements reported in the literature indicate silence at superficial lumbar muscles (Andersson et al., 1996; Dickey et al., 2003; Mathieu and Fortin, 2000; McGill and Kipper, 1994; Sarti et al., 2001; Solomonov et al., 2003; Toussaint et al., 1995). As for superficial thoracic extensor muscles at larger trunk flexion angles, some report continuation of activity (McGill and Kipper, 1994; Toussaint et al., 1995) while others suggest relaxation (Dickey et al., 2003; Mathieu and Fortin, 2000; Andersson et al., 1996) reported activity in deep lumbar muscles as well.

Abdominal muscles remain silent up to trunk flexion angles of about 115° at which angles global extensor muscles become inactive. Subsequently, abdominal muscles (especially internal oblique [IO], Figure 3.3) initiate activation up to the peak rotation, generating flexor moments that offset the moments produced by the passive component of back muscle forces. In other words, abdominals are activated to increase and maintain the large flexion angles. Activities in abdominal muscles have also been reported in earlier studies during full flexion as extensor muscles become silent (Mathieu and Fortin, 2000; McGill and Kipper, 1994; Olson et al., 2006).

The effect of changes in passive properties of muscles and ligamentous spine on the results in general and the FRP in particular is found to be substantial. A decrease in passive contribution of extensor muscles (case with +5 per cent) (Figure 3.6) markedly increased activity in global extensor muscles and diminished that in abdominal muscles at larger flexion angles. A reverse trend was computed when the passive contribution was increased resulting in an earlier and greater activity in abdominal muscles but flexion relaxation in extensor muscles. Similar effects were also predicted as the bending rigidity of the ligamentous spine was altered (Figure 3.7). A decrease in passive stiffness due to an injury or joint relaxation could delay flexion relaxation in extensor muscles. The abdominal muscles are also affected by such changes.

In conclusion, existing biomechanical models of the spine rarely take into account the equilibrium equations simultaneously at all directions and levels (Arjmand et al., 2006, 2007).
and, hence, risk violating essential equilibrium conditions at levels and directions different from the ones considered. Flexion relaxation in global thoracic and local lumbar extensor muscles at larger trunk flexion angles is a direct consequence of passive resistance of extensor muscles and ligamentous spine that both increase with the lumbar rotation. Alterations in these passive properties and in the relative lumbar-pelvic rhythm could, hence, influence the load redistribution and flexion relaxation phenomenon. Future Kinematics-driven model studies should, amongst others, account for asymmetry in movements and for system stability in the optimization procedure.

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References


