

Analysis of squat and stoop dynamic liftings: muscle forces and internal spinal loads

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Abstract Despite the well-recognized role of lifting in back injuries, the relative biomechanical merits of squat versus stoop lifting remain controversial. In vivo kinematics measurements and model studies are combined to estimate trunk muscle forces and internal spinal loads under dynamic squat and stoop lifts with and without load in hands. Measurements were performed on healthy subjects to collect segmental rotations during lifts needed as input data in subsequent model studies. The model accounted for nonlinear properties of the ligamentous spine, wrapping of thoracic extensor muscles to take curved paths in flexion and trunk dynamic characteristics (inertia and damping) while subject to measured kinematics and gravity/external loads. A dynamic kinematics-driven approach was employed accounting for the spinal synergy by simultaneous consideration of passive structures and muscle forces under given posture and loads. Results satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Net moments, muscle forces at different levels, passive (muscle or ligamentous) forces and internal compression/shear forces were larger in stoop lifts than in squat ones. These were due to significantly larger thorax, lumbar and pelvis rotations in stoop lifts. For the relatively slow lifting tasks performed in this study with the lowering and lifting phases each lasting ~2 s, the effect of inertia and damping was not, in general, important. Moreover, posterior shift in the position of the external load in stoop lift reaching the same lever arm with respect to

the S1 as that in squat lift did not influence the conclusion of this study on the merits of squat lifts over stoop ones. Results, for the tasks considered, advocate squat lifting over stoop lifting as the technique of choice in reducing net moments, muscle forces and internal spinal loads (i.e., moment, compression and shear force).

Keywords Muscle force · Finite element · Dynamic · Kinematics · Lifting technique

Introduction

Musculoskeletal impairments occur frequently and have a substantial impact on the health and quality of life of the population as well as on the health care resources. Search for a safer lifting technique has attracted considerable attention due to the high risk of injury and low back pain (LBP) associated with frequent lifting in industry. Compression force limits have been recommended for safer manual material handling (MMH) maneuvers based on the premise that excessive compression loads could cause injury. Despite the well-recognized role of lifting in low back injuries [4, 12, 17, 33, 53], the literature on safer lifting techniques remains controversial [25, 48]. In search of optimal lifting methods, squat lift (i.e., knee bent and back straight) is generally considered to be safer than the stoop lift (i.e., knee straight and back bent) in bringing the load closer to the body and, hence, reducing the extra demand on back muscles while counterbalancing the moments of external loads. The importance of the squat versus stoop lifting technique has, however, been downplayed due to the lack of a clear biomechanical

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rationale for the promotion of either style [25, 48, 66]. Many workers, despite instruction to the contrary, prefer the stoop lift due to its easier operation, lower energy consumption in repetitive lifting tasks [38, 42] and better balance [97]. Besides, it is known that squat lift is not always possible due to the lift set up and load size.

The advantages in preservation or flattening (i.e., flexing) of the lumbar lordosis during lifting tasks are even less understood. Lifting has been categorized as either squat or stoop often with no recording of changes in the lumbar lordosis, which may influence the risk of injury [68, 82]. The kyphotic lift (i.e., fully flexed lumbar spine) is recommended by some, as it utilizes the passive posterior ligamentous system (i.e., posterior ligaments and lumbodorsal fascia) to their maximum thus relieving the active extensor muscles [36, 37]. In contrast, however, others advocate lordotic and straight-back postures indicating that posterior ligaments cannot effectively protect the spine and an increase in erector spinae activities is beneficial in increasing stability and reducing segmental shear forces [22, 45, 47, 66, 100]. Moderate flexion has been recommended by model [6, 91, 92] as well as experimental studies [2] to reduce risk of failure under high compressive forces. As the lumbar posture alters from a lordotic one to a kyphotic one, the effectiveness of erector spinae muscles in supporting the net moment (due to smaller lever arms [50, 62, 99] and the anterior shear force (due to changes in line of action [68]) decreases while the passive contribution of both extensor muscles and the ligamentous spine increases [6, 8, 62].

Evidently, an improved assessment of various lifting techniques and associated risk of tissue injuries depends directly on a more accurate estimation of the load partitioning in human trunk in dynamic lifting conditions. The spinal loads are influenced not only by the gravity, inertia and external loads, but also, more importantly, by trunk muscle forces (due to their smaller moment arms and their compensatory response to stability demands and tissue injuries). Despite conflicting data in the literature, previous studies generally indicated a decrease in trunk strength but an increase in trunk moments, muscle coactivity, muscle forces and spinal loads as the trunk movement was performed at a faster rate [20, 27, 28, 39, 40, 41, 55, 69]. A number of optimization and EMG driven models have been used to estimate the muscle forces in various lifting conditions. Many of these models do not properly account for the nonlinear passive resistance and/or complex geometry/loading/dynamics (i.e., inertia and damping) of the spine [16, 26, 35, 93]. In addition, many are simplified in not considering dynamic equilibrium

equations simultaneously in all directions and at all levels.

Towards a more accurate estimation of muscle forces and spinal loads in lifting tasks, an iterative hybrid dynamic kinematics-based finite element approach is introduced and applied in which a priori measured kinematics of the spine along with nonlinear passive properties are exploited. The kinematics-based approach that results in a synergistic solution of the active/passive system has already been successfully applied to isometric conditions in upright [30, 89] and flexed postures [8]. The objectives of this work are, hence, set as follows.

1. To extend kinematics-based approach to dynamic conditions by accounting for the inertia and damping in the solution of nonlinear transient equations of motion over the lifting period,
2. To perform in vivo measurements of trunk kinematics, needed as input data in subsequent model studies, on subjects performing sagittally symmetric forward/backward lifting tasks with/without loads in squat and stoop techniques,
3. To evaluate muscle forces (active/passive components) as well as spinal loads at different levels under loading and kinematics considered in vivo while accounting for the curved path (i.e., wrapping) of global extensor muscles,
4. To assess effects of dynamic characteristics of trunk (inertia and damping) and positioning of the external load on results.

Materials and methods

In vivo measurement

Fifteen healthy men with no recent back complications volunteered for the study after signing an informed consent form approved by the Institute de réadaptation de Montréal. Their mean (\pm SD) age, body height and mass were 30 ± 6 years, 177 ± 7 cm and 74 ± 11 kg, respectively. While bending slightly forward, light-emitting diode, LED, markers were attached on the skin at the tip of the T1, T5, T10, T12, L1, L3, L5 and S1 spinous processes for evaluation of lumbar and torso flexions. Three extra LED markers were placed on the posterior/superior iliac spine and ilium (left/right iliac crests) for the evaluation of pelvic rotation, and one on the load to track the position of weights in hands. A three-camera Optotrak system (NDI International, Waterloo, ON, Canada) was employed to collect 3D coordinates of LED markers.

Subjects were asked, with no instruction on lumbar posture, to perform sagittally symmetric squat (knee bent) and stoop (knee straight) lifts with and without 180 N weight placed on a bar in front at 20 cm height from the floor. Each task lasted 4–5 s and started from upright standing with no load in hands and ended again in upright standing with or without load in hands. Two-way analyses of variance (ANOVA) for repeated measure factors were performed to study the effect of different lifting techniques (2 levels: stoop and squat) and load magnitude (2 levels: 0 and 180 N) on collected kinematics data (Statistica, StaSoft, Tulsa, OK, USA).

Thoracolumbar finite element model

A sagittally symmetric T1–S1 beam-rigid body model made of six deformable beams to represent T12–S1 discs and seven rigid elements to represent T1–T12 (as a single body) and lumbosacral (L1–S1) vertebrae was used [6, 8, 30, 89]. The beams modeled 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 loads along with flexion versus extension differences were represented in this model based on numerical and measured results of previous single- and multi-motion segment studies [6, 73, 80, 85, 88, 104]. The trunk mass and mass moments of inertia were assigned at gravity centers at different levels along the spine based on published data for trunk segments [77, 78] and head/arms [105]. Connector elements parallel to deformable beams were added to account for the intersegmental damping using measured values [51, 63]; translational damping = 1,200 N s/m and angular damping = 1.2 N m s/rad. For the cases with 180 N external load in hands, the inertia of the load was computed using measured kinematics and subsequently added as an external load.

Prescribed postures

Measured sagittal rotations at thorax (evaluated by the change in inclination of the line attaching the T1 marker to the T12 one) and pelvic (evaluated by the orientation of normal to the plane passing through the three markers on the pelvis) of one subject were applied at the T12 and S1 vertebrae, respectively. The total lumbar rotation, calculated as the difference between the foregoing two rotations, was subsequently partitioned in between various segments based on values reported in earlier investigations [3, 29, 34, 76, 79, 91, 104]. Relative proportions of ~7, 12, 15, 22, 27

and 17% were used to partition the lumbar rotation between various motion segments from T12 to L5 levels, respectively.

Muscle model and muscle force calculation

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles was used (Fig. 1) [6, 8, 10, 94]. In order to accurately simulate the curved path of global muscles (i.e., longissimus thoracis pars thoracic and iliocostalis lumborum pars thoracic) at flexion angles considered in this study, these muscles were constrained to follow a curved path whenever their distances from T12 to L5 vertebral centers at undeformed configuration diminished beyond ~10% (i.e., to reach the limit values of 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12–L5 vertebrae, respectively). This wrapping mechanism, similar to that formulated in our earlier works [86, 90, 92], was considered in order not to allow the line of action of these muscles reach unrealistically close to the vertebrae resulting in erroneous small lever arms at different levels that occurs in larger flexions when global muscles are simulated as straight lines between their insertion points. During the analysis, when a wrapping is detected at a vertebral level, the contact force between the muscle and the corresponding vertebra is evaluated using the equilibrium equation in the

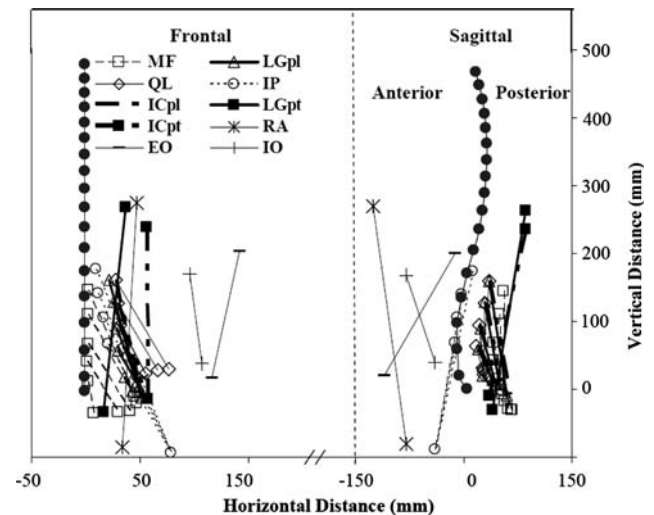


Fig. 1 Representation of the model as well as global and local musculatures in the sagittal and frontal planes. Fascicles on one side are shown; *ICpl* iliocostalis lumborum pars lumborum, *ICpt* iliocostalis lumborum pars thoracic, *IP* iliopsoas, *LGpl* longissimus thoracis pars lumborum, *LGpt* longissimus thoracis pars thoracic, *MF* multifidus, *QL* quadratus lumborum, *IO* internal oblique, *EO* external oblique and *RA* rectus abdominus

instantaneous configuration assuming a frictionless contact that results in a constant muscle force along its entire length. This wrapping contact force was considered as an additional external force in subsequent iteration.

To evaluate muscle forces, kinematics-based algorithm was employed to solve the redundant active/passive system subject to prescribed measured kinematics, inertia, damping and external loads. In this manner, calculated muscle forces at each instance of loading were compatible with the prescribed kinematics (i.e., posture) and loads while accounting for the realistic nonlinear stiffness of the passive system as well as trunk dynamic characteristics. Initially, the model calculated the moments at different levels required for the a priori prescribed rotations (i.e., measured posture). To resolve the redundancy problem at each level (i.e., in partitioning the calculated moment among muscles attached to that level), an optimization approach with the cost function of minimum sum of cubed muscle stresses was also needed along with inequality equations of unknown muscle forces remaining positive and greater than their passive force components (calculated based on muscle strain and a tension–length relationship [19]) but smaller than the sum of their respective maximum active forces (i.e., 0.6 MPa times muscle's physiological cross-sectional area, PCSA) [103] and the passive force components. Once muscle forces were calculated, the axial compression and horizontal shear penalties of these muscle forces along with wrapping contact loads (if needed) were fed back into the finite element model as additional updated external loads. This iterative approach was continued at each time instance till convergence was reached. The finite element program ABAQUS [1] was used to carry out nonlinear transient analyses while the optimization procedure was analytically solved using an in-house program based on Lagrange Multipliers Method [84]. Implicit algorithm with unconditionally stable Hilber–Hughes–Taylor [46] integration operator was used to solve the problem. The time step was automatically selected by the solver but was constrained to remain <0.01 s.

Parametric studies

In order to determine the relative role of inertia on results, the lifting case with squat technique and 180 N in hands was re-analyzed with both trunk and external load inertias neglected (i.e., quasi-static analysis). To further investigate the effect of inertia, another quasi-static analysis of the same task was

performed with the inertia of external load considered as an additional load. As for the effect of damping on results, additional dynamic analyses were performed assuming either a totally undamped system or one with fivefold increase in damping simulating an over-damped condition.

Subjects carried the 180 N load further away (anteriorly from their S1 level) in stoop lift than in squat lift with the mean difference of <88 mm. In order to examine the effect of such load positioning on our predictions, the stoop lift was re-analyzed with the load position in the horizontal direction shifted closer to the body (i.e., the S1 level) to become identical to that in squat lift.

Results

Subjects carried stoop lifts, as compared with squat, with significantly larger thorax, pelvis and lumbar (T12–S1) rotations ($P < 0.000002$, 0.05 and 0.03, respectively). The magnitude of load (0 vs. 180 N) did not, however, have any significant effect on these rotations. The temporal variations of pelvic and thorax rotations of one typical subject measured under four different cases along with the intervening lumbar vertebral rotations were used as input data into the subsequent model studies. Polynomials of sixth order were fitted on these rotations in order to smooth prescribed data ($R^2 > 98\%$, Fig. 2) into the model. Positions of the external load (180 N in hands) in the model studies of the squat and stoop lifts were based on the mean of measurements; the horizontal location of the load was nearly the same in both lifts when evaluated with respect to the T12 level, whereas it was more anterior (by <88 mm) in stoop lifts when calculated with respect to the S1 for each subject.

The maximum net external moment at the L5–S1 disc substantially increased as 180 N load was carried in hands and as the lifting was performed in stoop technique (by ~28% compared to squat lifts) (Fig. 3). These moments were carried primarily by muscles with a small contribution (~10–30% depending on the lifting technique) from the passive ligamentous spine. At the T12 level and under both loading conditions, the moments resisted by the global extensor muscles and ligamentous spine were both larger in stoop lifts than in squat lifts (Fig. 4). The contribution of active muscle forces, especially in case with 180 N in hands, was greater than that of the passive muscle forces. The relative differences in global muscle forces in squat versus stoop lifts were due primarily to the smaller passive components in the former lifts (Fig. 4). Maxi-

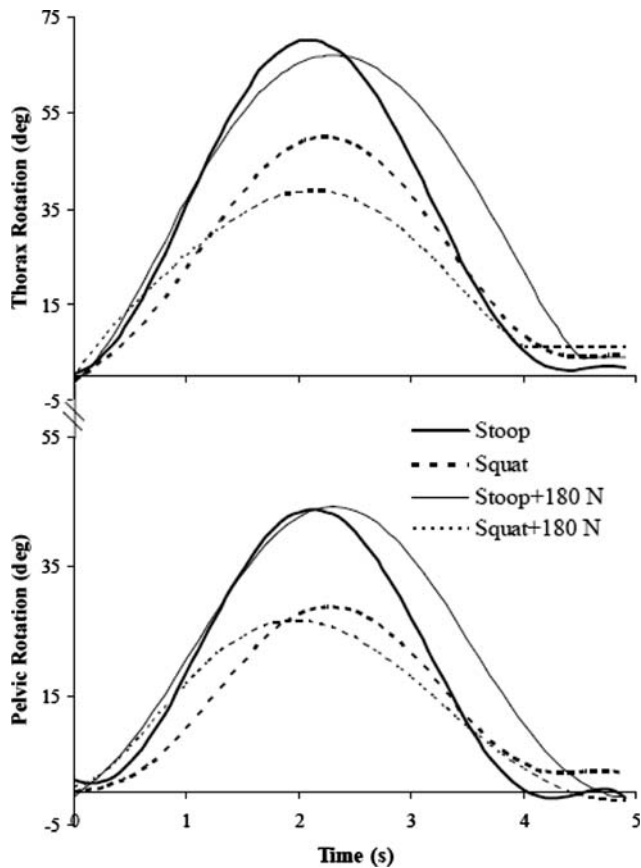


Fig. 2 Prescribed thorax (*top*) and pelvis (*bottom*) rotations in the model for various cases based on in vivo measurements of a typical subject (smoothed by 6th order polynomials, $R^2 > 98\%$). The T12–S1 rotations are subsequently prescribed in the model based on the difference between these two rotations and proportions given in the text

lum muscle forces at different local and global levels (Fig. 5) were larger in stoop lifts than in squat lifts. Internal local compression and shear forces at different intervertebral disc levels were also greater in stoop lifts than in squat lifts with maximum differences reaching ~800 N in compression and ~200 N in shear at the L5–S1 level (Fig. 6; Table 1). Calculated shear forces showed a dramatic increase from the L4–L5 level to the L5–S1 in all cases. Due to larger intersegmental lumbar rotations, passive segmental moments were also larger in stoop lifts (Table 1).

Except for the time periods at the beginning and end of tasks as well as immediately after lifting the external load, results were almost the same for both static and dynamic analyses over the entire duration of motion (Fig. 7). Inclusion of the inertia of the external load in the analysis was also found to have negligible effects on results. Increasing the damping at the motion segments did not change results, while

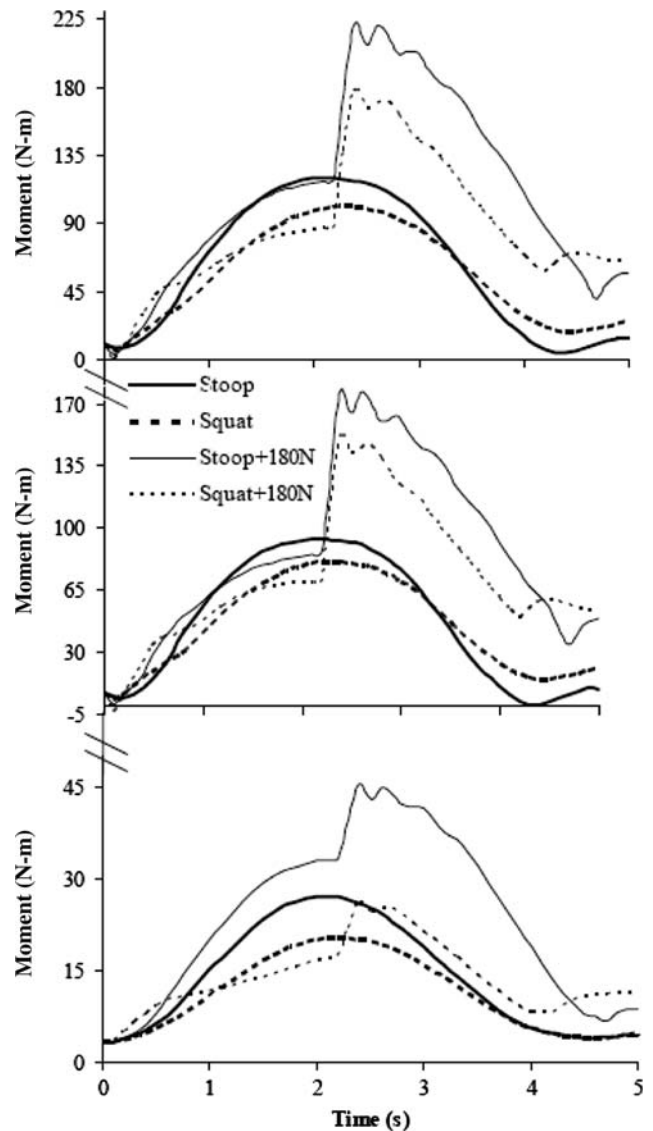


Fig. 3 Predicted temporal variation of sagittal moments at the L5–S1 level for different cases (N m); net external moment (*top*), portion resisted by muscle forces (*middle*) and portion resisted by passive ligamentous spine (*bottom*). For the cases with load in hands, the sharp increase in moments is noted as the load reaches its maximum value of 180 N in 0.2 s duration

considerable fluctuations in response (± 10 N m on required thorax moment) were noted in the absence of any damping in the model (Fig. 7). Closer positioning of the external load in stoop lift (by <88 mm in order to arrive at the same relative lever arm with respect to the S1 as that considered in squat lift) reduced the total muscle forces as well as net moment and internal compression/anterior shear forces at the L5–S1 level to values in between those predicted for squat and stoop lifts, i.e., nevertheless remaining greater than those for the squat lift.

Fig. 4 Predicted temporal variation of net external moment at the T12 level (*top*) and associated active (*middle*) and passive (*bottom*) global muscle (*LGpt* longissimus and *ICpt* iliocostalis) forces for different lifting techniques without any load in hands (*left side*) and 180 N load in hands (*right side*). The rising time of 180 N external load applied in hands is shown by *lines* on the *right*. *S* moment resisted by passive spine, *M* moment resisted by muscles

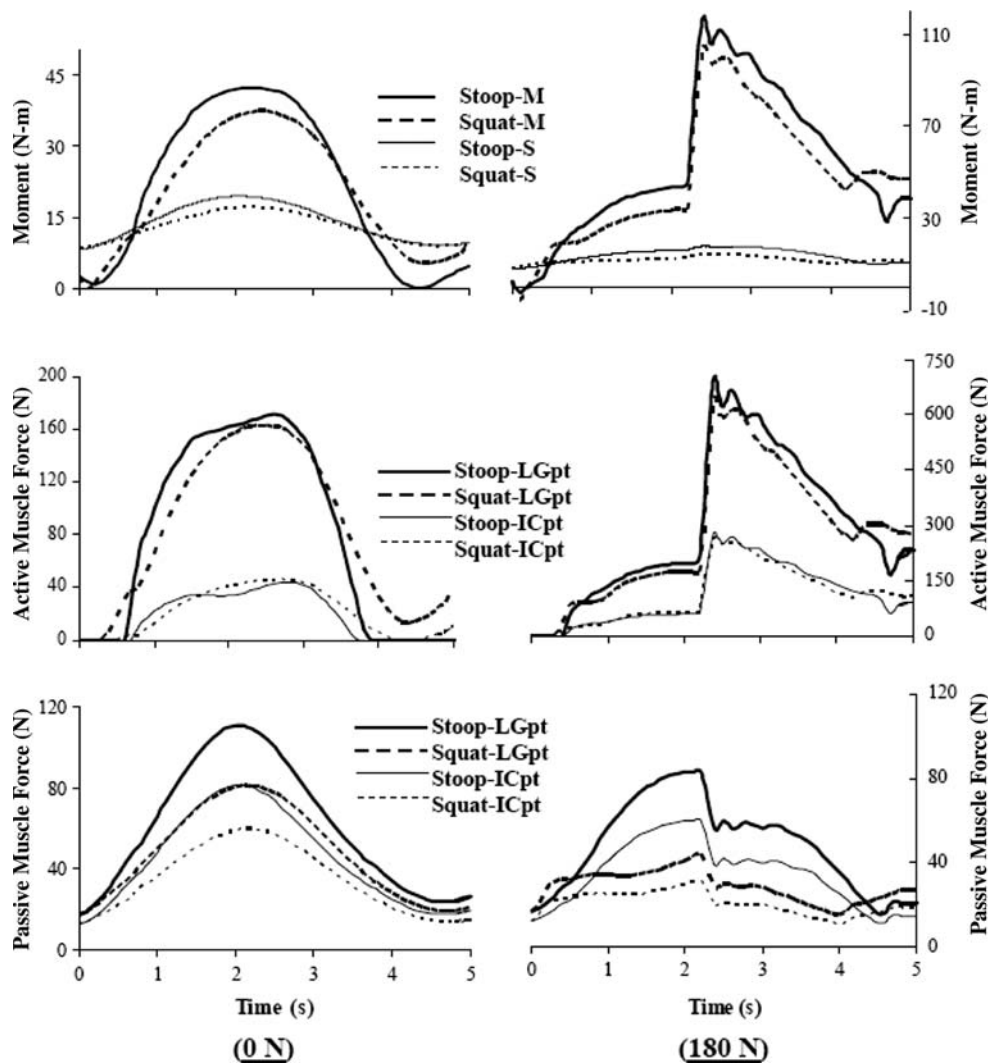
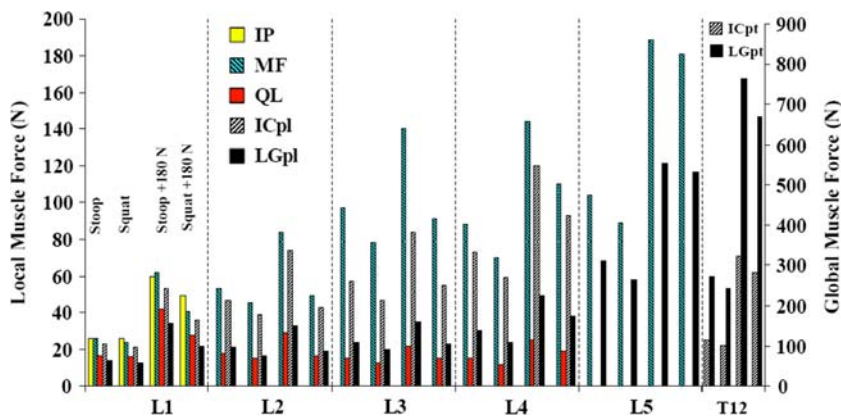


Fig. 5 Maximum predicted total local and global muscle forces at various levels for different cases



Discussion

The controversy on a safer lifting technique persists due partly to the complexity of the problem (e.g., dependence on changes in the posture, pelvic con-

straints and load positioning) and oversimplifications (assumptions involving kinematics, constraints, geometry, material properties, loading, dynamic characteristics, etc.) in model studies. In this work, the kinematics-based approach that has previously been

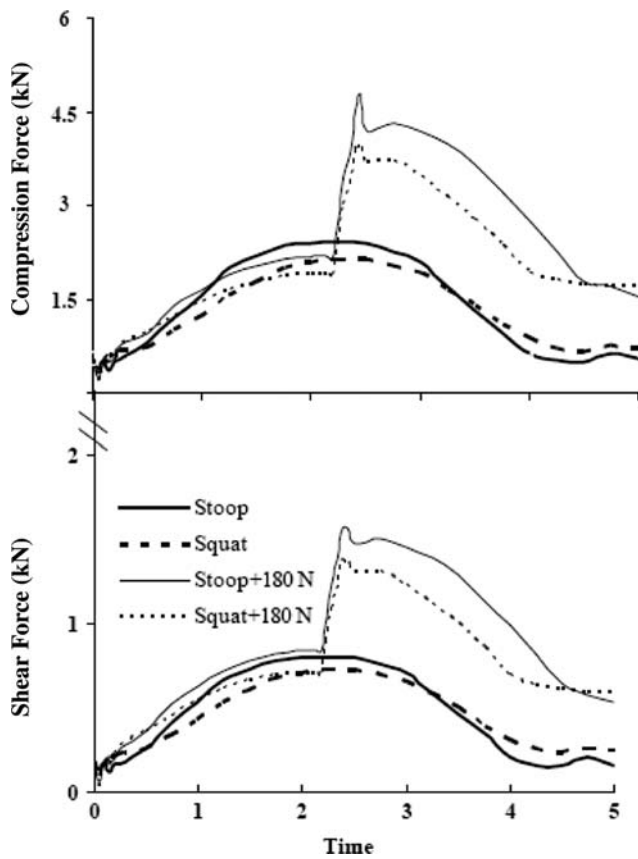


Fig. 6 Computed temporal variation of local compression (*top*) and anterior shear (*bottom*) forces at the L5–S1 level for different cases. These forces are normal and tangential to the disc mid-height planes

applied to isometric lifting conditions was extended to predict muscle forces and internal spinal loads in dynamic stoop and squat lifts. For this purpose, parallel in vivo studies were performed to collect kinematics required as input data into the model. The entire forward–backward movements were carried out over 4–5 s with either squat or stoop techniques but no instructions on the lumbar posture.

Methodological issues

Evaluation of the segmental rotations from skin markers is recognized to involve errors in the identification of vertebral positions, skin movement relative to the underlying vertebrae and deformability of vertebrae themselves [14, 56, 60, 87, 106]. Due to these inherent errors, the measurements were used to evaluate temporal variations of pelvic tilt and thorax rotation while the intervening lumbar segmental rotations were evaluated based on the partitioning of the difference between foregoing measured rotations using the relative values reported in the literature. The assumption of rigid body motion at the T1–T12 segments (upper torso) in the model was justified, in agreement with others [72, 96], by measuring nearly equal rotations for lines attaching either the markers T12–T5 or markers T12–T1. Changes in the relative proportions used to partition the total T12–S1 rotation among intervening segments would, as expected, alter to some extent the net moment, passive ligamentous resistance and muscle recruitments at these levels. Moreover, although these proportions were assumed constant during the entire lifting tasks, such may not necessarily be true in vivo as the relative demand at different levels could vary during lifting. These relative ratios were taken from data obtained in static measurements [29, 34, 76, 79], which have also been used in previous dynamic studies [70, 82, 83] in order to evaluate the contribution of passive tissue in offsetting external load. To prescribe measured rotations in the model, kinematics data of one typical subject rather than the mean of all subjects were considered. This was done due mainly to noticeable variations in the duration of lowering/lifting phases in between subjects.

The transverse abdominal, latissimus dorsi, lumbo-dorsal fascia, inter- and multisegmental muscles were neglected, whereas the oblique abdominal muscles were presented by straight single lines rather than

Table 1 Maximum internal loads in passive ligamentous spine for different cases at various levels; passive segmental moment, *M* (N m), local compression force, *C* (N), and local anterior shear force, *S* (N)

| Disc level | 0 N | | | | | | 180 N | | | | | |
|------------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|
| | Stoop | | | Squat | | | Stoop | | | Squat | | |
| | <i>M</i> | <i>C</i> | <i>S</i> | <i>M</i> | <i>C</i> | <i>S</i> | <i>M</i> | <i>C</i> | <i>S</i> | <i>M</i> | <i>C</i> | <i>S</i> |
| T12–L1 | 20 | 926 | 226 | 17 | 902 | 187 | 18 | 2,416 | 384 | 14 | 2,315 | 222 |
| L1–L2 | 24 | 1,155 | 244 | 19 | 1,121 | 196 | 21 | 2,921 | 381 | 14 | 2,660 | 192 |
| L2–L3 | 24 | 1,445 | 184 | 18 | 1,374 | 135 | 19 | 3,383 | 244 | 12 | 2,922 | 85 |
| L3–L4 | 20 | 1,793 | 300 | 14 | 1,675 | 249 | 15 | 3,903 | 536 | 7 | 3,274 | 376 |
| L4–L5 | 20 | 2,162 | 258 | 14 | 1,989 | 227 | 19 | 4,518 | 502 | 9 | 3,704 | 425 |
| L5–S1 | 23 | 2,355 | 800 | 16 | 2,159 | 737 | 33 | 4,831 | 1,635 | 16 | 4,023 | 1,416 |

Maximum internal loads occur nearly at the time of maximum trunk flexion

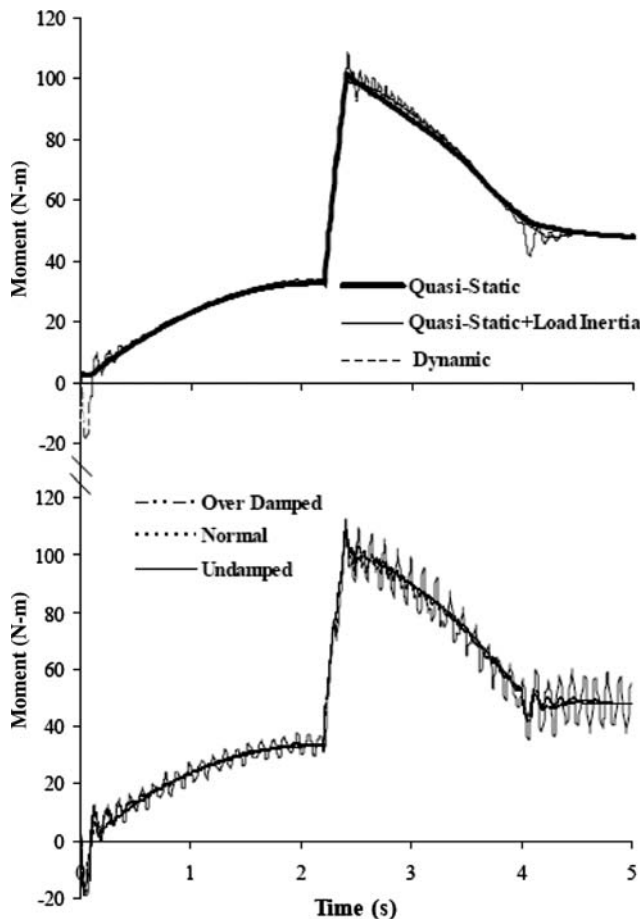


Fig. 7 Predicted effect of changes in system dynamics characteristics on the net moment at the T12 level for the squat lift with 180 N in hands; effect of consideration of trunk and load inertias (*top*) and of damping (*bottom*)

curved sheets of muscle. Consideration of several fascicles instead of just one for oblique muscles (EO and IO) has influenced the estimated spinal loads significantly in asymmetric lifting tasks but only slightly in symmetric ones [21]. Indirect effect of the transverse abdominal and latissimus dorsi muscles in unloading the spine through lumbodorsal fascia have been reported not being sizable during lifting tasks [11, 15, 61, 65, 70, 95]. Moreover, the likely mechanical effects of the intra-abdominal pressure (IAP), neglected in this study, have been found to depend on the posture and the co-activity level of abdominal muscles [7]. While local muscles were modeled as straight lines between their respective insertion points, realistic muscle paths were considered for global extensor muscles by wrapping them over all T12–S1 vertebrae whenever in the course of lifts their distance to associated vertebral bodies reduced more than 10% of their initial distances. This allowed for a maximum of ~10% reduction

in muscle lever arms at different levels during flexion which was chosen in accordance with published data in the literature [50, 62, 99]. The wrapping of global muscles occurred at all levels under larger flexion angles and resulted in curved paths with realistic lever arms at different levels. Had straight lines been assumed for global muscles, much smaller lever arms would have been generated resulting in greater muscle forces and internal loads. The wrapping contact forces (Table 2) remained relatively small compared with muscle forces suggesting minor changes in lines of action at wrapping points.

In the presence of nonlinearity in equations, numerical integration using an unconditionally stable implicit method was employed in this study. Minimum sum of cubed muscle stresses, as the cost function used in the optimization, has been recognized to agree better with EMG data [5, 24, 49]. The convergence of the nonlinear optimization solution on a global minimum was assured in this study by solving the optimization problem analytically. For the sake of comparison with EMG measurements of earlier studies, the computed muscle forces were partitioned, at the post-processing phase of the analysis, into passive and active components using a passive tension–length relationship for all muscles [19]. Moreover, the maximum allowable muscle stress of 0.6 MPa was assumed for all muscles neglecting the effect of activation level on this value. It is important to emphasize that the passive load–length relationship considered for muscles in the current study have absolutely no bearing at all on the predicted spinal loads and total muscle forces. The rate-dependent viscoelastic properties of the spinal segments, which could play a role at much higher loading rates, [71, 101, 102] were not considered in this study. Finally, in accordance with parallel *in vivo* measurements, the response was limited to the sagittal plane, thus neglecting out of plane motions.

Table 2 Maximum wrapping contact forces for different cases at various levels (N)

| Vertebra level | 0 N | | 180 N | |
|----------------|-------|-------|-------|-------|
| | Stoop | Squat | Stoop | Squat |
| T12 | 0 | 0 | 0 | 0 |
| L1 | 40 | 27 | 60 | 32 |
| L2 | 55 | 39 | 113 | 66 |
| L3 | 62 | 43 | 97 | 44 |
| L4 | 80 | 55 | 168 | 89 |
| L5 | 99 | 70 | 251 | 138 |

Maximum wrapping contact forces occur nearly at the time of maximum trunk flexion

Effect of dynamic parameters

Generally, faster trunk movements have been associated with a decrease in trunk strength but increases in trunk moments, muscle coactivity, muscle forces and spinal loads [20, 27, 28, 39, 40, 41, 55, 69]. Inertia effects of the trunk and external load have been indicated to play a noticeable role at the onset of a lift with jerky movements [55]. Our results showed a negligible effect of inertia forces on trunk moment and spinal loads except at three time intervals; the beginning and end of the tasks as well as a short period after picking the load up (Fig. 7) which agrees with earlier observations [47]. Apart from these periods, a quasi-static analysis would yield sufficiently accurate results with no real need to account for inertia forces which could be due to the slow lifting performed by our subjects (i.e., lowering and lifting periods each lasting ~2 s). Our results also demonstrated that the inertia of the trunk, and not that of the load, was the major factor for the observed differences in these three time periods. In a different lifting condition, however, the latter has been estimated to be more important than the former [69]. The computed net moment at the L5–S1 is noted to be in good agreement with values reported in previous dynamic studies [18, 32, 52, 96].

Although recognized as an important parameter, damping has been neglected in earlier biomechanical model studies of dynamic lifting [52]. A fivefold increase in the segmental damping value which was used in the model based on earlier measurements did not markedly alter predictions of this work, especially away from the three time intervals indicated earlier (Fig. 7). Introduction of damping appeared to primarily smooth the temporal response by removing high-frequency fluctuations (i.e., noise).

Effect of lifting techniques

The relative lumbar/pelvic rotations during lowering/lifting phases showed greater contributions in all cases from the pelvis than the lumbar spine (by as much as twofold) and remained within the range of data reported in the literature [31, 41, 64, 81]. Thorax and pelvis rotations were both larger in stoop lifts compared to those in squat lifts (Fig. 2) resulting in greater lumbar (T12–S1) rotations in stoop lifts by 10.5° and 5.9° in cases with and without 180 N load in hands, respectively. These additional flattenings of the lumbar spine in stoop lifts increased the wrapping contact forces (Table 2) and moment-carrying contribution of passive ligamentous spine and trunk muscles. Moreover, despite identical lever arms considered (based on

measurements) for the external load of 180 N at the T12 level, the net moments and hence muscle forces and internal loads were all greater in stoop lifts than in squat ones; e.g., maximum net moments of 200 N m and 160 N m were predicted at the L5–S1 level for stoop and squat lifting, respectively. Same trends were also found in the absence of external loads or even when the external load was shifted by <88 mm closer to the body in the stoop lift in order to reach the same lever arm with respect to the S1 as that considered in the squat lift.

Therefore, results of this study appear to suggest the squat lift as the safer lifting technique in reducing the net moment, muscle forces and internal ligamentous loads at all levels. It should be emphasized that the relative merits of these lifting techniques depend not only on the relative rotations at the thorax, pelvis and lumbar spine but also on other factors such as position of external loads, voluntary alterations in the lumbar curvature and speed of movement. These could partly be the reason why the literature remains yet inconclusive as some report smaller net moment and trunk load in squat lifting [13, 43, 58, 83] while others indicate otherwise [23, 28, 57, 59, 98]. The reduction in net moment in squat lifts, under all cases with and without external load, is due primarily to smaller pelvic and lumbar (and hence thorax) rotations in this technique resulting in much reduced net moments from the mass of the upper body and the external load about the L5–S1 (Fig. 8). Variations in the location of external loads and rotations of pelvis and lumbar spine from a lift to another, as expected in different studies, are important and could substantially influence the results and subsequent comparison of lifting techniques towards identification of the optimal one. The biomechanical advantages for the squat lifts in our study would become even more apparent had a smaller lever arm for the external load been considered in these lifts [9, 98]. In an earlier combined in vivo model study on the effect of changes in the lumbar curvature on trunk response in isometric lifts with identical thorax rotations [6], the maximum segmental shear/compression forces and activity in extensor muscles occurred in the lordotic posture while the maximum segmental flexion moment occurred in the kyphotic posture. The kyphotic postures exploited primarily the passive ligamentous/muscle force components while the active muscle forces played more important role in lordotic postures. The study advocated the free style posture or a posture with moderate flexion as the posture of choice in static lifting tasks when considering both internal spinal loads and active/passive muscle forces. One must note that in that study the thorax rotation remained nearly

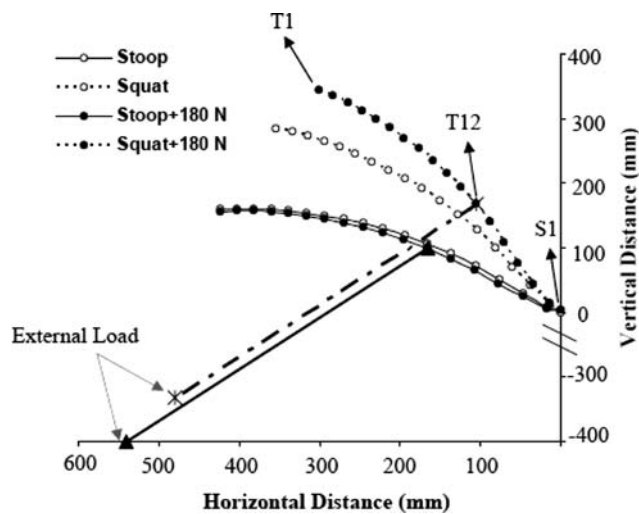


Fig. 8 Deformed configurations of the model at the beginning of lifting phase (i.e., end of lowering phase) for various cases. The position of the external load held in hands, also shown, has identical horizontal lever arms with respect to the T12 in both squat and stoop configurations. The deformed configurations have been shifted in both horizontal and vertical directions to place the S1 at the origin of axes. The thorax rotation is much larger in stoop lifts (70° and 66.9° without and with load, respectively) than in squat lifts (49.7° and 38.4°)

the same irrespective of changes in the lumbar curvature. In the current study, however, the thorax rotations of 66.9° and 70° in stoop lifts, respectively, with and without 180 N in hands were much greater than corresponding rotations of 38.4° and 49.7° in squat lifts (Fig. 8). Although we did not investigate the effect of changes in lumbar curvature in dynamic stoop and/or squat lifts, the conclusions of the previous isometric study advocating a flattened lumbar spine and current dynamic one advocating a squat lift (involving more lordotic lumbar curvature) do not contradict each other due to the crucial effect of posture (i.e., thorax and pelvic rotations) on results. Earlier studies on the effect of posture in lifting have suggested a lordotic posture in increasing the extensor activity during the early phases of the lift [22, 45, 100].

Results of previous works on extensor muscle activities in stoop lifts usually demonstrate two peaks: the first and smaller one occurring in lowering phase while the second and larger one in lifting phase of the tasks [44, 54, 67, 74, 75]. Our predictions on active extensor muscle forces also show similar variations during the tasks (Fig. 4). Due to the relatively small flattening of the lumbar spine (T12–S1) considered in the model (remaining $<26^\circ$), no flexion relaxation was observed which would otherwise have influenced the results in the final periods of the lowering phase of the study.

In conclusion, the current work while accounting for nonlinear properties of the ligamentous spine, wrapping of global extensor muscles, trunk dynamic characteristics (inertia and damping) and in vivo measured postures, calculated muscle forces and internal spinal loads during squat and stoop lifts using a novel dynamic kinematics-based approach. The model accounted for the spinal synergy by simultaneous consideration of passive ligamentous structure and muscle forces under given posture and loads. The predictions, therefore, satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Results, for the tasks considered, advocate squat lifting over stoop lifting as the technique of choice in reducing net moments, muscle forces and internal spinal loads. These values remained greater, though to a lesser extent, even when the lever arm of the external load in stoop lift was reduced to become equal to that in squat lift. These were due to significantly larger thorax, lumbar and pelvis rotations in stoop lifts. Furthermore, for the relatively slow lifts performed and modeled in this work, dynamic characteristics of trunk did not demonstrate significant effects on results.

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