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The ratio of thoracic to lumbar compression force is posture dependent

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Despite the evidence suggesting that between 8% and 55% of manual labourers experience thoracic pain, research on spinal loading during occupational tasks has been almost invariably limited to the lumbar spine. In this study, we determined the ratio of thoracic to lumbar compression force and the relative risk of injury to each region in various postures. Compressive forces on the spine were calculated based on previously reported thoracic and lumbar intradiscal pressures and disc cross-sectional areas. Flexion postures were associated with an approximate doubling in lumbar compression force but only small increases (or even decreases) in thoracic compression. The ratio of thoracic to lumbar compression was above the tolerance ratio (i.e. the ratio of thoracic to lumbar compressive strength) during upright postures and below the tolerance ratio during flexion postures, indicating that upright postures may pose a greater relative risk of injury to the thoracic spine than to the lumbar spine.

Practitioner summary: Previously reported thoracic and lumbar *in vivo* disc pressures during various postures were compared. The ratio of thoracic and lumbar compression increased during upright postures and decreased in flexed postures, indicating that upright postures may pose a greater risk of injury to the thoracic spine than to the lumbar spine.

Keywords: thoracic spine; lumbar spine; spinal compression; biomechanics; manual material handling

1. Introduction

Although not to the same degree as the lumbar spine, pain in the thoracic spine can be a significant source of functional limitation and work loss. A comprehensive review of reports on thoracic pain in the adult working population found prevalence rates between 8% and 55% in manual labourers (including workers performing manual material handling tasks) (Briggs et al. 2009a). One study of 217 workers who performed at least 25 lifts per day found a 33% prevalence of upper back symptoms, compared with a 58% prevalence of lower back symptoms (Yeung et al. 2002). In addition, acute disc herniations in the thoracic spine have occurred during lifting activities (Logue 1952; Mellion and Ladeira 2001; Polga et al. 2004; Vanichkachorn and Vaccaro 2000). Although the aetiology of pain and injury of the thoracic spine is poorly understood, mechanical loading may be one factor contributing to onset, severity and duration of symptoms (Briggs et al. 2009b). Despite a large percentage of the working population experiencing thoracic pain, research on spinal loading during occupational tasks has been almost invariably limited to the lumbar spine.

Employers often instruct their employees to lift with squat lifts (knees bent and back upright), as opposed to stoop lifts (knees straight and back bent), even though there is no evidence that this prevents low back injuries (Burgess-Limerick 2003). For example, even though squat lifts are often thought of as the ‘correct’ way to lift, the majority of studies have found lumbar compressive forces and moments during squat and stoop lifts to be within 5% of each other (Karwowski and Marras 1999; Straker 2003). Although biomechanical models of the lumbar spine have been extensively used to understand the loading on lumbar spine from such postures, loading on the thoracic spine is less well understood and efforts at biomechanical modelling have been limited (Iyer et al. 2010; Thaxton 2009; Wilson 1994). Furthermore, anatomical and biomechanical differences between the thoracic and lumbar spine, including the contribution of the ribcage and sternum, complicate modelling of the thoracic region. To ensure that postures that are thought to be protective of the lumbar spine do not inadvertently increase loads on the thoracic spine, it is important to gain a better understanding of the loading on the thoracic spine.

Intradiscal pressure measurement is the only direct way to determine the *in vivo* loading conditions in the human spine (Nachemson and Morris 1964; Sato, Kikuchi, and Yonezawa 1999; Schultz et al. 1982). There have been numerous intradiscal pressure studies conducted on the lumbar spine over the past several decades (Nachemson and Morris 1964; Sato, Kikuchi, and Yonezawa 1999; Schultz et al. 1982; Wilke et al. 2001). By contrast, there has been only one study on the intradiscal pressure in the thoracic spine (Polga et al. 2004). In addition, recorded intradiscal pressures have been used in combination with the cross-sectional area of the discs to calculate the compressive force on the lumbar spine (Brinckmann

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and Grootenboer 1991; Sato, Kikuchi, and Yonezawa 1999), but not on the thoracic spine. Therefore, the objective of this study was to determine the ratio of thoracic to lumbar compression force during various postures using the above method and previously reported data. This should provide insight as to how compressive forces imparted to the thoracic spine differ from the lumbar spine.

2. Methods

2.1 Intradiscal pressure in the thoracic spine

Polga et al. (2004) published data on the intradiscal pressures within the thoracic spine. Specifically, the intradiscal pressure of one middle (T6/7 or T7/8) and one lower thoracic disc (T9/10 or T10/11) was taken from four male and two female volunteers, with a mean age of 28 years (range, 19–47) and a mean body mass of 73 kg (range, 54–81 kg). The measurements were taken using custom pressure-measuring needles from Gaeltec Ltd (Dunvegan, Isle of Skye, Scotland). Once the pressure-sensing device was in place, the authors recorded measurements while the volunteers assumed various standing postures, with and without hand loads. For example, in one posture, subjects stood upright with 20 kg in their hands and elbows flexed to 90°. In another posture, subjects flexed their torso 30° from upright with 20 kg in their hands and arms straight down. Although Polga et al. (2004) collected data during a total of 18 postures, including many sitting postures, we limited this study to the data from eight standing postures (Table 1). In addition, since thoracic disc herniations typically occur in the lower thoracic spine (Maiman and Pintar 1992; Mellion and Ladeira 2001; Vanichkachorn and Vaccaro 2000), we only used data from T9/10 and T10/11.

2.2 Compressive forces on the thoracic spine

The mean stress of an ideal mechanical system is simply the applied force divided by the area perpendicular to the applied force, meaning one could calculate compressive force by multiplying the measured pressure by the cross-sectional area. As has been experimentally both shown and modelled by Brinckmann and Grootenboer (1991), the pressure in the nucleus pulpous is not equal to the axial force on the disc divided by the area since the disc bulges during compression (Brinckmann and Grootenboer 1991; Nachemson 1960). The radial bulging imposes a tensile stress in the axial direction that is superimposed on the stress from the applied compressive force. To accurately determine the axial force applied to the disc, a correction factor is needed that accounts for the difference between the intradiscal pressure and the mean stress. Based on empirical data from thoracolumbar discs, the ratio of intradiscal pressure to mean stress is 1.54 ± 0.12 [mean ± standard deviation (SD)] (Brinckmann and Grootenboer 1991; Nachemson 1960). The correction factor of 1.54 indicates that about 65% of the total measured pressure (1/1.54) is due to the applied compressive force, while 35% (0.54/1.54) is due to the tensile stress associated with radial bulging. Using this known correction factor, we calculated the compressive force on the disc, F_{comp}, according to Equation (1), where the intradiscal pressure, P, is multiplied by the cross-sectional area of the disc, A, and divided by 1.54.

$$F_{\text{comp}} = \frac{\mathrm{P} \times \mathrm{A}}{1.54}. \tag{1}$$

Table 1. Mean previous measurements of *in vivo* pressures in the thoracic and lumbar spine.

Task	Thoracic spine	Lumbar spine	
	Data from Polga et al. (2004) (MPa)	Data from Sato, Kikuchi, and Yonezawa (1999) (MPa)	Data from Wilke et al. (2001) (MPa)
Holding 20 kg while upright and elbows flexed 90°	2.30		1.10
Holding 20 kg with flexion of 30° and arms straight down	1.40		2.30
Upright while holding 20 kg and arms at the sides	1.43		
Upright	0.86	0.54	0.50
Flexion of 30°	1.09	1.32	0.95
Extension of 15°	1.00	0.60	0.60
Twist of 30°	1.07		0.70
Lateral bending	1.15		0.59

Although the cross-sectional areas of the discs from the volunteers in Polga et al.'s (2004) investigation were not reported, the mean cross-sectional areas at T9/10 and T10/11, from individuals with the same mean age (29 ± 9 years), have been reported elsewhere as 10 ± 1.8 cm^2 (Koeller, Meier, and Hartmann 1984). Representative data for the cross-sectional area from individuals of the same age are required for our analysis since the cross-sectional area of intervertebral discs increases with age. Given the same mean age and disc levels, we believe that the reported data on the cross-sectional area of thoracic discs from Koeller, Meier, and Hartmann (1984) can be used in combination with the *in vivo* pressure measurements from Polga et al. (2004) to determine the spinal compressive forces imposed on the thoracic spine.

2.3 Thoracic versus lumbar compression force

For comparison with the compression in the thoracic spine, the compressive forces in the lumbar spine were calculated based on previously published studies on the *in vivo* pressures in the lumbar spine. Although intradiscal pressures have been studied as far back as the 1960s and 1970s (Nachemson 1960; Nachemson and Morris 1964; Sato, Kikuchi, and Yonezawa 1999; Schultz et al. 1982), these earlier studies used pressure transducers mounted into flexible needles that could easily be bent for postures other than that of upright standing. This may have introduced significant artefacts (Polga et al. 2004). Therefore, we relied on two studies (Sato, Kikuchi, and Yonezawa 1999; Wilke et al. 2001) that used comparable transducer technologies to those used by Polga et al. (2004). Sato, Kikuchi, and Yonezawa (1999) recorded the L4/5 intradiscal pressures from eight individuals during three postures that were comparable to those used by Polga et al. (2004) (Table 1). The mean age of the participants was 25 years (range, 22–29 years) and the mean mass was 73 kg (range, 60–96 kg). Wilke et al. (2001) recorded the L4/5 intradiscal pressures for one individual, a 45-year-old male with mass of 70 kg, during seven comparable postures (Table 1). Although the postures across the three previous investigations were not specifically designed to be equivalent, we were, in this study, able to establish the similarity of each posture investigated based on the description of the postures along with accompanying photographs and figures. For instance, Wilke et al. (2001) presented the lumbar disc pressure as a function of trunk flexion angle (Figure 2 of that study). The value associated with 30° of flexion was compared to the thoracic disc pressure presented in Polga et al. (2004) for 30° of trunk flexion. Also, Wilke et al. (2001) showed a photograph of the subject holding 20 kg with the elbows flexed to 90° (Figure 3(d) of that study and described as holding 20 kg close to the body in that study), the exact posture described and tested in Polga et al. (2004). A similar comparison was made for each of the seven postures evaluated in this study.

Given that the lumbar pressure from the one subject from the study of Wilke et al. (2001) was within the range of pressure from the eight volunteers from the study of Sato, Kikuchi, and Yonezawa (1999) for upright standing, flexion of 30° and extension of 15° (Table 1), we believe it is reasonable to combine the two lumbar pressure data-sets to conduct our analysis. The biggest discrepancy between these two postures was flexion of 30°, with the mean pressure from the study of Sato, Kikuchi, and Yonezawa (1999) of 1.32 MPa, compared to 0.95 MPa from the study of Wilke et al. (2001). However, the lowest recorded pressure from the study of Sato, Kikuchi, and Yonezawa was 0.90 MPa, or below the recorded pressure from the study of Wilke et al. (2001).

The mean L4/5 disc area from the two lumbar intradiscal pressure studies was 16 ± 1.7 cm^2 (Sato, Kikuchi, and Yonezawa 1999; Wilke et al. 2001). Based on the intradiscal pressure and disc area data, we calculated the compressive forces on the lumbar spine using Equation (1). Although the intradiscal pressures on the lumbar and thoracic spine were measured on different subjects, the overlap in postures allowed a direct comparison to be made of the compressive forces on the thoracic and lumbar spine by normalising compressive forces to body weight (BW). Finally, to determine the relative risk of injury between the two regions of the spine, we compared the ratio of thoracic to lumbar compression force to the tolerance ratio, defined as the compressive strength of the lower thoracic spine divided by the compressive strength of the lumbar spine, of about 0.88 (Yamada 1970).

2.4 Comparison to other biomechanical models

To assess the accuracy of our estimates of thoracic compression force, we compared the normalised compressive forces to the forces predicted by an optimisation-based thoracic biomechanical model of the spine (Iyer et al. 2010). Specifically, we evaluated the per cent difference between our results and the results of the optimisation-based model, for the postures of standing upright, flexion of 30°, extension of 15°, holding 20 kg with trunk flexion of 30° and arms straight down and holding 20 kg while upright and elbows flexed 90°.

Similarly, to assess the accuracy of our estimates of lumbar compression force, we compared the normalised compressive forces calculated using Equation (1) with those predicted by the University of Michigan Three-Dimensional Static-Strength Prediction Program (3DSSPP, Ann Arbor, MI, USA). This model is the most widely used computer software program in ergonomics (Dempsey, McGorry, and Maynard 2005). Specifically, a 50th percentile male was modelled using 3DSSPP, in the postures of standing upright, flexion of 30°, extension of 15°, holding 20 kg with trunk flexion of 30° and arms straight down and holding 20 kg while upright and elbows flexed 90°. The resulting compressive forces from 3DSSPP were normalised to BW and then compared with the normalised compressive forces calculated in our study.

3. Results

3.1 Compressive forces on the thoracic spine

Holding 20 kg while upright and the elbows flexed imposed a compressive force on the lower thoracic spine of 2.1 times BW (Figure 1). Comparatively, holding the same weight with the arms straight down and the trunk flexed 30° generated 40% lower compressive force on the thoracic spine.

During upright stance with no external load, the compressive force on the thoracic spine was 0.78 times BW. Adding a 20-kg load at the hands with the same posture increased the compressive force by 67%. Compared to upright stance (no external load), torso flexion increased the compression force imparted to the thoracic spine by 27%. By contrast, when holding a 20-kg load, changing posture from upright (with arms by the side) to torso flexion of 30° (with arms hanging straight down) changes the compressive force in the thoracic spine by less than 5%. In addition, holding 20 kg and flexing to 30° increased thoracic compressive force by 28%, compared with the same posture (torso flexion) and no external load.

3.2 Ratio of thoracic to lumbar compression force

The ratio of thoracic to lumbar compressive forces varied with posture, ranging from 0.32 while holding 20 kg and flexing 30° with the arms straight down to 1.12 while holding 20 kg while upright with the elbows flexed (Figures 1 and 2). The higher ratio while holding the weight with an upright posture was due to a 40% decrease in the thoracic compressive force and a 109% increase in lumbar compressive force. The ratio of thoracic to lumbar compression force while holding 20 kg and flexing 30° with the arms straight down (0.32) was 64% below the tolerance ratio of 0.88. The ratio of thoracic to lumbar compression force while holding 20 kg while upright and the elbows flexed (1.12) was 27% above the tolerance ratio.

With no hand loads, a change in posture from standing upright to torso flexion of 30° caused a 140% increase in the compressive force in the lumbar spine (compared with only a 27% increase in the thoracic spine), resulting in thoracic/lumbar compression ratios of 0.99 and 0.52, respectively. Adding 20 kg of external load while in a flexed posture lowered the ratio even further to 0.32, due to a 105% increase in lumbar compressive force and only a 30% increase in thoracic compressive force.

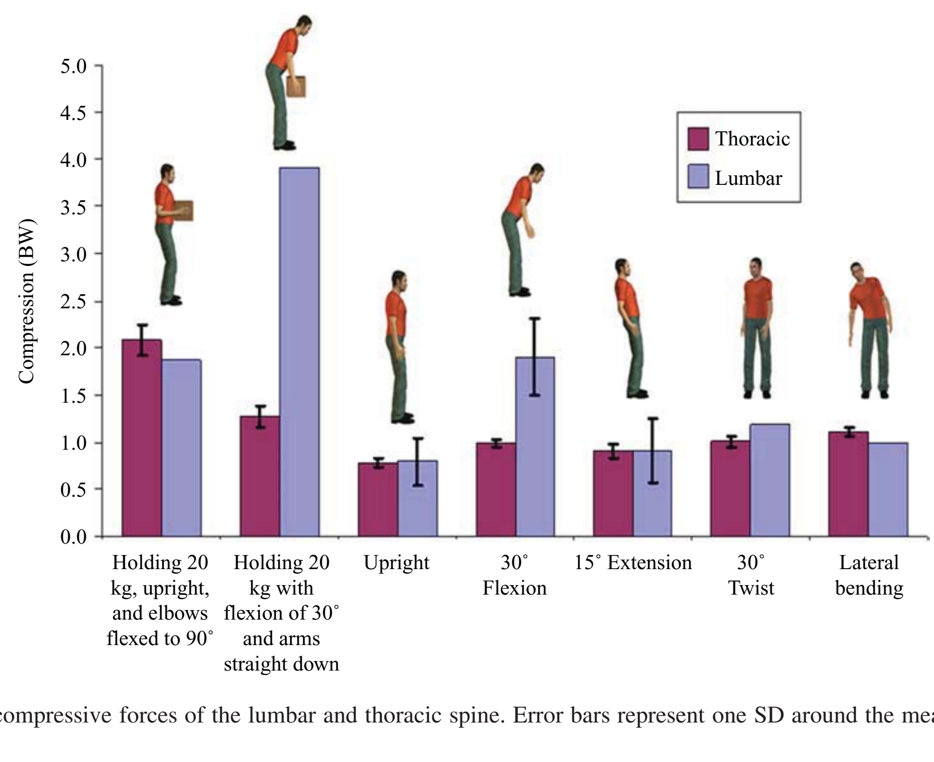

Figure 1. Mean compressive forces of the lumbar and thoracic spine. Error bars represent one SD around the mean.

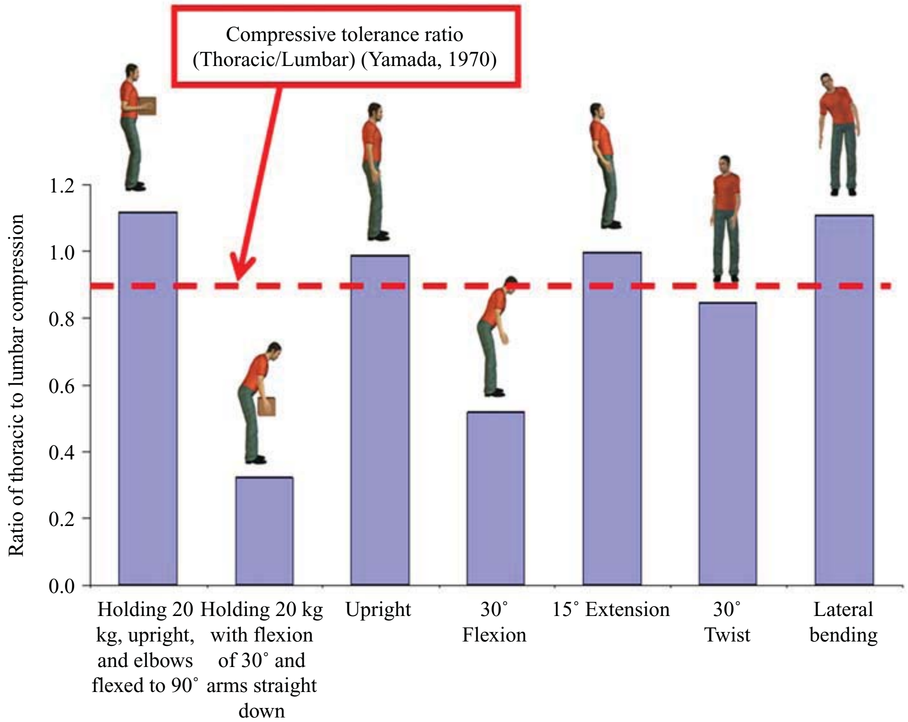

Figure 2. Ratio of thoracic to lumbar compression force to the ratio of the compressive strength between the lower thoracic and lumbar spine of about 0.88 (Yamada 1970) (indicated as a dashed red line). For postures involving an upright trunk, the ratio of thoracic to lumbar compression is above 0.88, indicating an increased risk of injury. The ratio is well below 0.88 for postures involving trunk flexion.

There was little variation between the thoracic/lumbar compression ratios during standing upright, extension of 15°, twist of 30° and lateral bending, with a mean value of 0.95 ± 0.12 times BW. The ratio of thoracic to lumbar compression forces during these postures was slightly above (8%) the tolerance ratio of the thoracic to lumbar spine.

3.3 Comparison to other biomechanical models

The compressive forces on the thoracic spine calculated based on intradiscal pressure were strongly correlated with the forces calculated by an optimisation-based biomechanical model ($r^2 = 0.97$), though the compressive forces calculated by

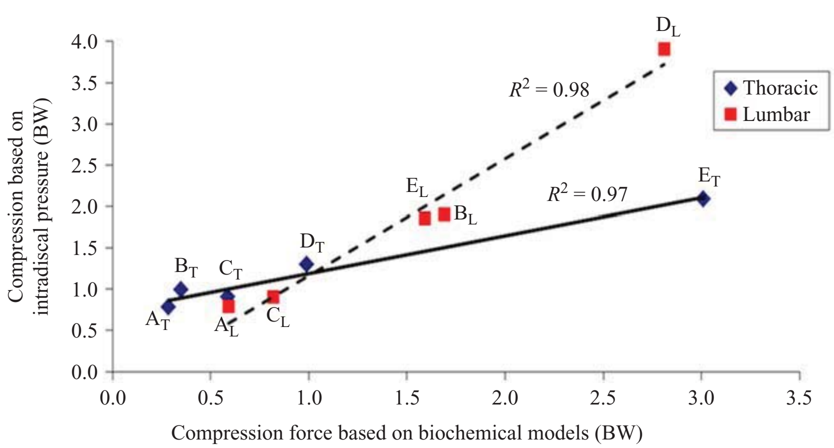

Figure 3. Compressive forces based on *in vivo* intradiscal pressure were compared with those predicted by a thoracic biomechanical model (Iyer et al. 2010) (blue diamonds, solid line) and a lumbar biomechanical model (3DSSPP) (red squares, dashed line) for various postures: (A) upright, (B) 30° flexion; (C) 15° extension, (d) holding 20 kg with flexion of 30° and arms straight down and (E) holding 20 kg while upright and elbows flexed 90°.

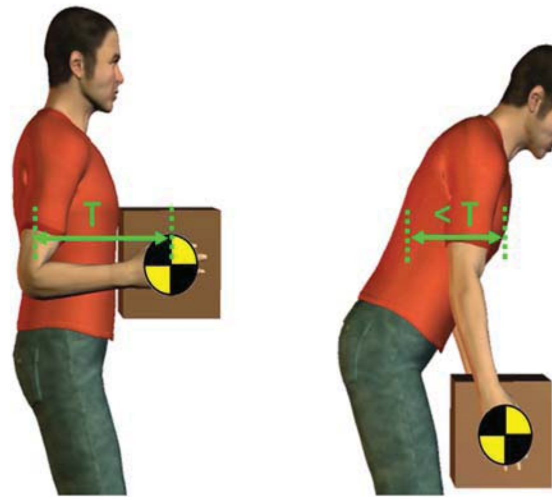

Figure 4. A change in posture from holding an object while standing upright to flexing decreases the external moment arm of the thoracic spine, *T*. For example, for a 50th percentile male, *T* decreases by 45%.

the optimisation-based biomechanical model were on average 31 ± 40% less than those calculated here (Figure 3). The biomechanical model underpredicted every posture, except holding 20 kg while upright with the elbows flexed.

The compressive forces on the lumbar spine calculated based on intradiscal pressure were strongly correlated with the force calculated by 3DSSPP ($r^2 = 0.98$). The compressive forces predicted by 3DSSPP were on average 18 ± 8% less than those calculated here. The 3DSSPP model underpredicted the compression force in every posture, compared with the intradiscal pressure calculations.

4. Discussion

Our main purpose was to determine the relationship between thoracic and lumbar compression force during a variety of standing postures. Based on previous studies that measured the intradiscal pressure during static postures (Polga et al. 2004; Sato, Kikuchi, and Yonezawa 1999; Wilke et al. 2001) and an empirical relationship between measured intradiscal pressure and compressive force (Brinckmann and Grootenboer 1991; Nachemson 1960), we found that the ratio of thoracic and lumbar compression force is not uniform and is highly dependent on the posture (Figure 2). In particular, we found that flexion postures are associated with an approximate doubling in lumbar compressive force but only small increases (or even decreases) in thoracic compression. Upright postures may pose a greater relative risk of injury to the thoracic spine than to the lumbar spine, given that the ratio of thoracic to lumbar compression force was above the tolerance ratio during upright postures and below the tolerance ratio during flexion postures.

Our results are comparable to a previous optimisation-based biomechanical model of the thoracic and lumbar spine that predicted an increase of 180% in lumbar spine compression, but only a 25% increase in thoracic spine compression after flexing 30° from an upright posture (Iyer et al. 2010). Iyer et al. also evaluated the effects of external hand weights, with both a 30° flexed posture (arms straight down) and an upright posture with the elbows bent. Using 10 kg total weight, the ratio of thoracic to lumbar compression with 10 kg hand weight was about 0.28 in the flexed posture and 0.98 in the upright posture with the elbows bent (Iyer et al. 2010). Although the amount of external weight is different from that used in our study, the trend is similar, with a significantly lower ratio during the flexion posture.

Our results are also consistent with fundamental statics. One of the main factors that influence the compressive force imposed on the spine during manual material handling is the external moment arm, meaning that the horizontal distance (in the sagittal plane) from the spine to the centre of gravity of the object is being lifted (Marras et al. 1995). The external moment arm for the thoracic spine decreases during trunk flexion since the object can be directly under the chest (Figure 4). For example, based on anthropometric software (HumanCAD v1.1, NexGen Ergonomics, Quebec, Canada), for a 50th percentile male, the external moment arm for the thoracic spine decreases 45% with the torso flexed 30°. Interestingly, the external moment arm for the lumbar spine remains unchanged. Another factor that influences the compressive force is the internal moment arm, meaning that the horizontal distance (in the sagittal plane) is from the spine to the centre of mass of the head, arms and torso (HAT). In flexed postures, the HAT internal moment arm increases substantially for the lumbar spine and only moderately for the thoracic spine. The change in internal and external moment arm with trunk flexion yields a smaller thoracic to lumbar compression force ratio compared with upright postures. There was a minimal change in internal moment arm (in the sagittal plane) from the spine to the centre of mass of the HAT during upright standing, extension of 15°, twist of 30° and lateral bending, likely explaining why the thoracic compression force and the ratio of thoracic to lumbar compression force during these postures were very similar to one another. Interestingly, the mean compressive force ratio during upright standing, extension of 15°, twist of 30° and lateral bending, of about 0.95, is consistent with the ratio of compressive strength between the lower thoracic and lumbar spine of about 0.88 (Yamada 1970). Thus, for these postures, there does not appear to be a substantial injury risk trade-off in either the thoracic or lumbar spine.

Although the postures were previously recorded during static holds, holding 20 kg while upright and the elbows flexed is a potential posture used during a squat lift. Similarly, holding 20 kg and flexing 30° with the arms straight down are a potential posture employed during a stoop lift. Given the myriad of object weights and sizes lifted in the workplace, it is possible that the ratio of thoracic and lumbar compression force may differ under alternate postures used during squat and stoop lifts from these postures than the one combination that simulated squat and stoop postures. However, based on these two postures, a squat lift may increase the compression and therefore the relative risk of injury to the thoracic spine, compared to a stoop lift. Further investigation on the ratio of thoracic to lumbar compression force, using more postures and object weights associated with manual material handling, is warranted.

How to design jobs and instruct employees to perform their jobs in order to prevent back injuries has been debated for decades (Garg and Herrin 1979; Straker 2003; Van Dieen, Hoozemans, and Toussaint 1999). Although the rule of thumb has been that the optimal posture allows the individual to lift with the object as close to the body as possible (Karwowski and Marras 1999; Luepongsak et al. 1997; Marras et al. 1995), in an attempt to decrease low back injuries, most employers instruct their employees to lift with squat lifts (Burgess-Limerick 2003). However, there is no indication that squat lifts decrease the risk of low back injuries (Burgess-Limerick 2003; Karwowski and Marras 1999). Previous studies have also demonstrated that other anatomic regions are placed at an increased risk of injury during occupational squat lifts compared to stoop lifts. For example, hip joint contact pressures double during occupational squatting compared to stooping, given the same magnitude of hand forces (Luepongsak et al. 1997). Moreover, occupational tasks involving squatting are associated with increased rates of hip and knee osteoarthritis (Coggon et al. 1998, 2000). Given that our results indicate that postures involving the upright trunk may also increase the risk of injury to the thoracic spine, the investigation adds to the body of knowledge that there are trade-offs between postures used to perform a task and no posture is without risks (Karwowski and Marras 1999; Luepongsak et al. 1997).

Unlike models of the lumbar spine, models of the thoracic spine must account for the mechanical contribution of the ribcage and sternum. The ribcage and sternum add stiffness to the thorax and serve to reinforce the thoracic spine during compression and bending. Over the past 20 years, a few models of the thoracic spine have been developed, each with increasing sophistication. Wilson (1994) estimated the compression in the thoracic spine from T8 through T12 by assuming the ribcage bore weight such that the stress at thoracic levels was equal to the stress at L3. The anatomy for the model formulation was taken from a single male cadaver. Using more detailed anatomy from numerous magnetic resonance imaging and computed tomography studies, Thaxton (2009) also estimated compression in the thoracic spine. Most recently, an optimisation-based thoracic biomechanical model of the spine (Iyer et al. 2010) was used to determine the forces on the thoracic spine. The ratio of compressive force on the spine to the ribcage and sternum force was determined by modelling the structures as parallel springs in a mechanical circuit and the known stiffness of the ribcage and sternum from experimental data. Although there was a high correlation between the force estimated from that model and the results of this study ($r^2 = 0.97$), the mean underestimate of 31% by the model may be due to the nature of the optimisation routine used in this model (Iyer et al. 2010). The difference between *in vivo* compression and optimisation-based compression estimates may be a function of most objective functions and constraints minimising co-contraction (i.e. activation of muscles that oppose the desired movement) in such models despite sizable *in vivo* levels of co-contraction. For example, the optimisation-based model of the thoracic spine minimised the sum of cubed muscle intensities required to balance the external moments and forces, thereby ignoring or significantly underestimating co-contraction of the anterior muscles during demanding tasks that are necessary to stabilise the trunk. Similar optimisation-based models of the neck and low back also underestimate the forces on the spine by up to 35% (Bean, Chaffin, and Schultz 1988; Choi 2003; Choi and Vanderby Jr. 2000; Moroney, Schultz, and Miller 1988; Schultz et al. 1982). The developers of the thoracic model noted that their model likely underestimated the *in vivo* co-contraction of the torso (Iyer et al. 2010). Our results support that statement. The discrepancy between our results and the biomechanical optimisation model may also be due, in part, to the uniform intradiscal pressure to mean stress ratio used in Equation (1). Since thoracic discs bulge less than lumbar discs, due to their smaller height (Koeller, Meier, and Hartmann 1984), it is possible that the intradiscal pressure to mean stress ratio may vary along the spine. Despite this, the relative differences between postures would remain constant across each region of the spine. As such, going from an upright to a flexed posture would still cause a large increase in the lumbar spine and only a small increase (or even decrease) in the thoracic spine.

There are several limitations to our study. First, the comparison between thoracic and lumbar compression force in this study was based on data taken from different volunteers. Although the postures were substantially equivalent, minor postural differences may have introduced additional variability. For instance, using telemeterised vertebral body replacement (VBR), changes to arm posture from the arms hanging down to placing the arms on the thighs reduced the compressive force on the spine by 13% (Dreischarf et al. 2010). Unfortunately, the estimated compressive force on the thoracic spine from this study cannot be compared to estimates from telemeterised VBR since compression at the treated level is shared by not only the VBR but also the posteriorly inserted internal fixation device, the preserved part of the fractured vertebra and the attached bone graft. Therefore, data from telemeterised VBR do not represent the total compression force and are likely underestimates of the compression in normal individuals (Rohlmann et al. 2007). For example, based on telemeterised VBR data, the compressive force in the lumbar spine was reported as 1.1 times BW during torso flexion, while based on our results it is about 1.9 times BW (Figure 1). Our estimation of compression force was solely based on disc pressure and therefore is not an estimation of the total compressive force on the segment since we did not determine the compressive force on the facet joints. Previously, the compressive force on the facet joints as a function of total spinal compression force has been experimentally determined to be 22.3 $\pm$ 2.4%, 25.1 $\pm$ 5.1% and 30.4 $\pm$ 7.1% (mean $\pm$ sd) during flexion, neutral and extension postures, respectively (Yang and King 1984). Given that the total spinal compression force is the summation of disc compression and facet compression, a reliable determination of the total spinal compression could be made.

Although a known limitation of our study was that the pressure measurements were taken from static postures, and thus do not account for the effect of dynamic loading, the use of static analysis is consistent with the ergonomics literature. In fact, the most common and widely used analysis tools in the ergonomics community such as the NIOSH Lifting Guide and the 3DSSPP also do not account for the effects of dynamic loading. Nonetheless, determining the dynamic load on the thoracic spine clearly warrants further investigation. Finally, we have only considered one loading mechanism, spinal compression force, without determining the bending moment or shear forces on the spine (Adams and Hutton 1982; McGill 1997), even though bending and shear are thought to contribute to the risk of injury. However, since only one of the three intradiscal pressure studies (Sato, Kikuchi, and Yonezawa 1999) presented horizontal disc pressure, we were unable to evaluate the shear force on the spine.

In order to confirm our results, it would be advantageous to record pressures in both spinal regions simultaneously and to obtain concurrent disc cross-sectional area measurements for those volunteers. Given that the data used for this study were taken from different subjects and that our main result was the ratio of thoracic of lumbar compression force, this precluded us from performing a statistical analysis. Nevertheless, given that we have normalised the data from the thoracic and lumbar spine to BW and used disc cross-sectional areas from comparably aged volunteers, we believe that our results would be validated by further investigation. In addition, for four of the seven tasks investigated (Table 1), there were lumbar intradiscal pressure data for only one subject. However, because this subject's lumbar compression force during the remaining three tasks was within the range of compression from the other eight volunteers, we have confidence in the results. Moreover, the lumbar spine data on the two postures with hand weights came from the same study, which adds to our confidence of comparing these two postures.

For this study, we focused on the lower thoracic spine, since that is the region where traumatic injuries typically occur. Since intradiscal pressure data are available for the middle thoracic spine, similar comparisons can be made between the compression in the middle thoracic spine and the compression in both the lower thoracic and lumbar spine. Although an in-depth investigation of this issue is outside the scope of this study, a brief analysis indicated that, across all tasks, the compression in the middle thoracic spine averaged 68% of the lower thoracic compression, consistent with the tolerance of the middle thoracic spine, which is 67% of the lower thoracic spine tolerance (Yamada 1970). In addition, our study focused on standing postures. However, jobs that require sitting, especially those that require forward flexion (e.g. dentists, dental hygienist and sonographers), are associated with high rates of low back disorders (Valachi and Valachi 2003; Village and Trask 2007). Given that there is evidence of increased lumbar intradiscal pressure during sitting compared to standing (Sato, Kikuchi, and Yonezawa 1999), this issue may warrant future investigation.

5. Conclusions

Our study demonstrates that the ratio of thoracic to lumbar compression force is not uniform but instead varies with posture. Specifically, the compressive force imposed on the thoracic spine increased during holding a 20-kg weight with an upright trunk and the elbows flexed, as compared to holding the same weight with the arms straight down and the trunk flexed 30°.

Given that employers often instruct their employees to lift with their knees bent and back upright, as opposed to knees straight and back bent, even though there is no justification that this prevents low back injuries (Burgess-Limerick 2003), our results indicate that tasks performed with the back upright may increase the risk of injury to the thoracic spine. While the trade-offs between thoracic and lumbar compression force still need further investigation, we suggest that allowing employees to perform their tasks with a stoop or semi-stoop posture may decrease the risk of injury to the thoracic spine.

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