Journal of Electromyography and Kinesiology 24 (2014) 114–119

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Corrective sitting strategies: An examination of muscle activity and spine loading

Raquel Castanharo [a], Marcos Duarte [b], Stuart McGill [c,*]

[a] Neuroscience and Behavior graduate program, University of Sao Paulo, Sao Paulo, Brazil
[b] Biomedical Engineering, Federal University of ABC, Santo Andre, Brazil
[c] Department of Kinesiology, University of Waterloo, Waterloo, Canada

ARTICLE INFO

Article history:
Received 11 August 2013
Received in revised form 28 October 2013
Accepted 2 November 2013

Keywords:
Sitting posture
Spine loading
Electromyography

ABSTRACT

The purpose of this study was to quantify the load on the lumbar spine of subjects when they are asked to adjust from a slouched sitting posture into an upright posture with one of three different strategies: "free" (no instruction) and two coached patterns: "lumbopelvic" dominant and "thoracic" dominant. The activity of selected muscles and kinematic data was recorded from 20 volunteers while performing the three movement patterns to adjust sitting posture. Moments and forces at the lumbar spine were computed from an anatomically detailed model that uses kinematics and muscle activation as input variables.

The lumbopelvic pattern produces less joint moment on the lumbar spine (on average 31.2 ± 3.9 N m) when compared to the thoracic pattern (43.8 ± 5.8 N m). However, the joint compression force was similar for these two patterns, but it was smaller in the free pattern, when no coaching was given (lumbopelvic: 1279 ± 112 N, thoracic: 1367 ± (125 N, free: 1181 ± 118 N). Lower thoracic erector muscle activity and higher lumbar erector activity were measured in the lumbopelvic pattern in comparison with the other two. In summary the lumbopelvic pattern strategy using predominantly the movement of anterior pelvic tilt results in smaller joint moments on the lumbar spine and also positions the lumbar spine closest to the neutral posture minimizing passive tissue stress. This may be the strategy of choice for people with low back flexion intolerance.

1. Introduction

The link between sitting posture and pain has increasingly become a topic of interest as the number of people adopting a sedentary lifestyle grows. Sitting generates a prolonged flexed posture of the lumbar spine (Endo et al., 2012), which is commonly associated with the development of low back disorders (McGill, 2007). This posture is associated with increased intradiscal pressure (Nachemson, 1981; Wilke et al., 1999), elevated disc degeneration (Videman et al., 1990), higher disc herniation rates (Wilder et al., 1988), and higher compression forces compared to standing (Callaghan and McGill, 2001). While sitting, one commonly adopts a large variety of postures and in particular when one wants to adjust or correct his or her posture from a slouch position.

Posture determines passive tissue stress, and the sharing of the stress between supporting tissues. Scannell and McGill (2003) found the lowest passive tissue stress when standing and when sitting upright as the spine curves and gets closer to its elastic equilibrium. When moving from a slouch position towards an upright trunk position, two basic movement patterns of the trunk in the vertical direction can be observed: one involving movement driven by hip flexion resulting in predominantly lumbar motion, and another involving movement predominantly at the thoracic lumbar junction emphasizing extension in this region of the spine (O'Sullivan et al., 2006).

A person who stays in the sitting position for long periods adjusts the trunk frequently. The way this is performed is important because creates different loads on the spine. Vergara and Page (2002) showed that people who work sited change the lumbopelvic position (movements bigger than 5°) every 6 min, on average, and if the average time interval between two consecutive changes is less than 5 min the probability of having lumbar pain is high. O'Sullivan et al. (2006) observed that the thoracic dominant pattern of movement for adjusting the posture from a slouch position to an upright trunk position is associated with more activation of muscles than the lumbopelvic pattern of movement. This finding suggests that the thoracic pattern of movement would be the less appropriate for some people; however, the compressive loads on the spine were not measured.

Not coincidentally, the thoracic pattern of movement is observed more often in a clinical examination than the lumbopelvic

* Corresponding author. Address: Department of Kinesiology, Spine Biomechanics Laboratory, University of Waterloo, 200 University Avenue West, Waterloo, Ontario N2L3G1, Canada.

E-mail address: mcgill@uwaterloo.ca (S. McGill).

1050-6411/$ - see front matter © 2013 Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.jelekin.2013.11.001

pattern of movement in patients with low-back pain (LBP), who show less ability to move their lumbar-pelvic region when sitting than people without pain (Dankaerts et al., 2006). These have been thought to be associated with some classifications of back pained people having stiffer hips (McGregor and Hukins, 2009; Mellin, 1990). The actual motions of the spine, hips, and pelvis and the corresponding spine load for each of these two movement patterns are yet to be determined. Knowledge of these biomechanical variables would assist in matching of the postural correction technique with the tolerances and pain mechanism of a person with a specific spine disorder.

The main purpose of this research study was to quantify the load on the lumbar spine of subjects when they are instructed to adjust the sitting posture either by adopting the thoracic or the lumbopelvic patterns, as well as an unconstrained free movement selected by the subjects without any coaching. It was hypothesized that the lumbopelvic pattern of movement would impose less load on the spine, specifically lesser joint compression and joint moment. It was also hypothesized that the lumbopelvic pattern of movement would generate less lumbar flexion and less activity of the thoracic erector spinae muscles.

2. Methods

2.1. Participants

Twenty one volunteers from the university community took part in the study (11 males – mean (SD), age 23 (3) years, height 1.80 (0.06) m, mass 78 (4) kg; 10 females – mean (SD), age 22 (3) years, height 1.64 (0.06) m, mass 62(7) kg). They formed a healthy sample with no history of disabling previous or current back pain, or musculoskeletal disorder. All participants read and signed the informed consent approved by the University Office of Research Ethics Board (ORE).

2.2. Tasks

The experiment involved each participant performing three different movement patterns while sitting, in a random order. They began by sitting on a stool without any backrest, with their feet on the ground, knees flexed approximately 90°, and their hands resting on their thighs (Fig. 1).

Participants performed the first pattern, called the "free" pattern, where they were asked to sit slouched and then correct the posture to sit upright. For the purpose of observing a natural pattern, the only specific instruction given was to slouch, and to do so in a way where the trunk did not bend forward excessively. The two coached patterns followed in a randomized order. In the first coached pattern, the "lumbopelvic" pattern, participants sat slouched and then were instructed to sit upright moving the pelvis in an anterior tilt, with an emphasis on hip flexion. The natural lumbar curve resulted from the pelvic motion. In the second coached pattern, called the "thoracic" pattern, participants sat slouched and then were instructed to sit upright focusing on extension through the thoracic lumbar transition region of the spine, by "lifting the ribcage". The lumbopelvic and thoracic patterns were demonstrated followed with a practice period prior to data collection. Each pattern was executed five times, from which the mean behavior was calculated.

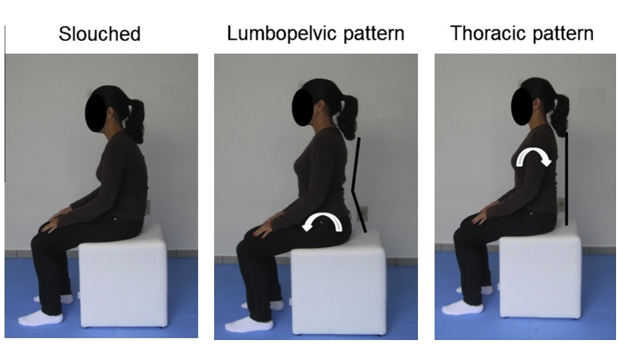


Fig. 1. The initial slouch position and the two coached patters.

2.3. Data collection

The dependant variables directly measured from participants were muscle activity, lumbar spine and pelvic angles, while joint moments and forces were computed from an anatomically detailed model that uses kinematics and muscle activation as input variables and it will be presented later. Ground reaction forces were also measured in a calibration procedure for the model.

2.4. Electromyography (EMG)

Surface EMG for quantifying muscle activity was obtained in the following way. The participants' skin was prepared to reduce the impedance to the myoelectric signal by shaving, rubbing the skin with an abrasive gel (Nuprep) and cleaning with a 50/50 ethanol/water solution. The EMG signal was recorded using bipolar, Ag–AgCl surface electrodes, placed with a center-to-center spacing of 2.5 cm over the following muscles, bilaterally, according to McGill (1992): rectus abdominis (3 cm lateral to the umbilicus), external oblique (approximately 15 cm lateral to the umbilicus), internal oblique (approximately midway between the anterior superior iliac spine and pubic symphysis, above the inguinal ligament), latissimus dorsi (lateral to T_9 over the muscle belly), thoracic erector spinae (5 cm lateral to T_9 spinous process), lumbar erector spinae (3 cm lateral to L_3 spinous process). Activity in the rectus femoris, a hip flexor (over the muscle belly 50% on the line from the anterior superior iliac spine to the superior part of the patella) and in the gluteus medius (approximately 4 cm below the iliac crest, posteriorly at the pelvis) was also recorded on the right side.

For normalization purposes all participants performed a maximum voluntary contraction (MVC) task for each muscle group, which consisted of a maximum isometric contraction against a research assistant's manual resistance according the procedure documented by Moreside et al. (2007). Specifically, the task for the abdominal muscles was executed in a seated position with the trunk inclined back approximately 60°. The participants were braced by the assistant and produced an isometric flexion contraction, followed by simultaneous right and left rotation and lateral bending efforts. For the back muscles, the Biering-Sorensen position was used: the participants were prone and cantilevered over a treatment table with the anterior superior iliac crests at its edge, while they extended their trunk to the horizontal position and hold this posture against a resistance. A pull up trial was conducted to reach the latissimus dorsi MVC, with the assistant holding the participant's body down, so as to ensure isometric contraction. For the gluteus medius, a hip abduction with the participants lying on their side was executed, and for the rectus femoris they were asked to perform a knee extension first and then a hip flexion while holding the knee extension. Mental focus on target muscles was coached throughout. A quiet lying trial in supine and prone positions was recorded to determine the noise bias.

The raw EMG signals were differentially amplified with a gain of 1000, common mode rejection ratio of 115 dB at 60 Hz, input impedance 10 GΩ by an EMG system (Model AMT-8, Bortec Biomedical, Calgary, Canada) and A/D converted at a sample rate of 2160 Hz (Vicon MX, Oxford Metrics, Oxford, UK).

2.5. Kinematic and kinetics

The participants' movements were registered with a three-dimensional motion analysis system: eight cameras, sampled at 60 Hz (Vicon MX, Vicon Motion Systems, Oxford, UK) operating with infrared light. In order to describe the body segments' movements, single reflective markers and markers on rigid clusters were attached on the body segments (Fig. 2). The thigh segment was defined by the greater trochanters, knee epicondyles and mid-thigh and a marker cluster was used in the middle of the segment. The origin of the coordinate system was in the greater trochanter. The pelvis was defined by the anterior superior iliac spines, iliac crests and pelvic depth, with a cluster fixed in the sacrum region. The origin of the coordinate system was the mid-point between the iliac crests which represented the middle of the L4/L5 disc. The trunk was defined by the left and right acromion, iliac crests and the anteroposterior pelvic radius, and a cluster was attached at the T12 level. The origin of the coordinate system was the mid-point between the acromion markers. All coordinate systems were adjusted to the long axis of the segment providing anatomical angles – for example flexion/extension, lateral bend and axial twist for the L4/L5 representing the lumbar spine.

A calibration trial was then performed capturing muscle activation and spine joint movement so that measured and modelled moments could be balanced with a EMG to muscle force gain factor that was then utilized for subsequent sitting trials. For this purpose the participants performed a squat (approximately 30°, 60° and 20° of knee, hip and lumbar flexion, respectively) holding a 20-kg bar while standing on two force plates (AMTI, 2160 Hz), with one foot on each one.

Signals obtained directly from the participants (force plate, EMG and the kinematics data) were collected simultaneously using Vicon software.

2.6. Data analysis

The EMG signals were filtered with a band-pass Butterworth filter of 4th order and zero lag at 30–500 Hz. To mimic the frequency response of the torso muscles' forces, the EMG signals were then full-wave rectified and low-pass filtered at 2.5 Hz (Brereton and McGill, 1998). Any noise offset bias was removed at this stage using the signal from quiet lying trials and each muscle signal was normalized by the MVC level and expressed as a percentage of MVC.

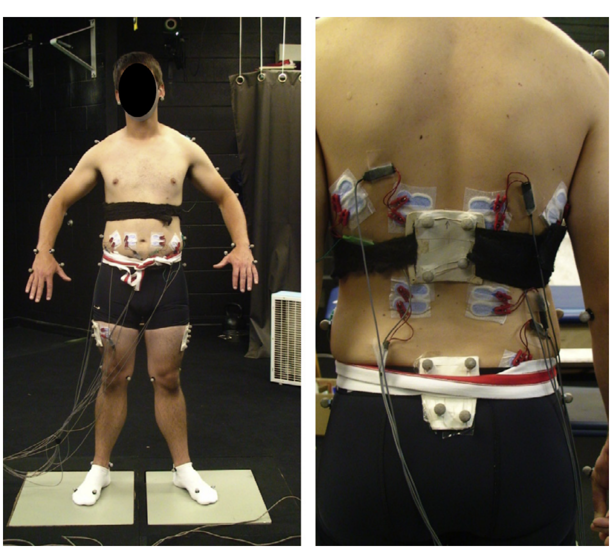


Fig. 2. Markers, clusters and electrodes' placement.

The reflective marker trajectories were filtered with a low-pass filter at 6 Hz with a Butterworth filter of 4th order and zero lag and then used to construct a linked segment skeletal model in Visual 3D software (C-motion Inc., Rockville, USA), which computed joint centers from the movement calibration and marker setup trials, from which subsequent joint angles, and joint reaction moments and forces were calculated. Thus, the reaction forces resulted from the gravitational and inertial force vectors from the upper body segments. Reaction moments were subsequently balanced with the restorative moments provided by the torso musculature. The lumbar movement was defined as the relative movement between the pelvis and the ribcage.

A lumbar spine model, that includes trunk muscles and passive tissues, was used to calculate the joint moment due to restorative forces and the compressive forces on the lumbar spine as the sum of reaction forces and forces from the muscles that span the joint (McGill, 1992, with updated anatomy documented in Cholewicki and McGill, 1996; Grenier and McGill, 2007). In this study the lumbar joint moment and force was calculated for the L4/L5 joint. The dynamic model is anatomically detailed, and uses EMG to estimate the force in 104 muscles, and tissue strain relationships to obtain passive forces in the ligaments and intervertebral discs (the interested reader is directed to a more complete description provided in (Cholewicki and McGill, 1996; McGill, 1992; McGill and Norman, 1986). Some muscles did not have direct EMG access. For example, the psoas muscle force was estimated using the EMG signal obtained from the rectus femoris (shown to be a reasonable surrogate for hip flexion moment prediction by Cholewicki and McGill, 1996). In this way the model recognizes individual muscle co-contraction patterns unique to person.

First, muscle force estimates for each individual (obtained from EMG, muscle size and geometry) were scaled using a least square fit by comparing the measured joint moment obtained from the calibration trial and the estimated moment computed from the first run of the spine model. This tuning of the model accommodated the differences in relative force producing ability of the many muscles used by each individual. Thus the tuning consisted of an iterative process such that the muscles were uniformly scaled so that their moment sum equalled the measured moment from the calibration trial. All subsequent analyses for a particular participant used their particular "muscle gain" obtained from this process. The L4/L5 joint compression force was calculated as the sum of the muscle compressive force from the spine model together with the reaction forces created by the masses of the upper body segments. Here a top-down inverse dynamic approach, including head, arms and trunk was facilitated with the Visual 3D software using the same joint centers and axis orientation previously described.

The time-series for the sitting tasks were normalized from 0% to 100% of the time required to complete the postural shifting movement. The lumbar and pelvic angles were normalized in amplitude by the maximum flexion range of motion of each individual so that larger lordosis angles (lumbar extension) correspond to smaller percentages of flexion, and larger anterior pelvis tilts correspond to greater percentages of flexion. The movements to set the spine and pelvis maximum range of motion were performed with the participants in a standing position. For this reason, the slouch sitting position did not started with exactly 100% lumbar flexion and 0% anterior pelvic tilt. Joint moment, joint compression and lumbar angle values were reported for the final upright sitting postures. Further, in order to describe the movement from slouch to upright, the mean joint moment and joint compression of the final 30% of the movement was calculated (called mean joint moment and mean joint compression). This is because the passive tissue contribution from ligaments in the fully flexed initial postures could not be accurately accounted for in this experiment. Other variables selected for analysis were thoracic and lumbar erector spinae maximum muscle activity together with the activation levels when fully upright at the end of the movement.

2.7. Statistical tests

Normality of scores was confirmed with Kolmogorov–Smirnov tests. A one-way repeated measures ANOVA with three levels (free, thoracic and lumbo-pelvic pattern), was performed followed by a Bonferroni post hoc test for pairwise comparison. A significance level of 0.05 was used for all statistical tests.

3. Results

All the participants adopted the same slouch posture in the beginning of each movement strategy (65% of lumbar flexion, on average). The lumbar flexion spine posture in the final upright position was larger (i.e. closest to elastic equilibrium) in the lumbopelvic pattern, followed by the free pattern and then the thoracic pattern (Table 1). The time series of the mean lumbar angle of all subjects is shown in Fig. 3. The thoracic pattern resulted in the largest mean joint moment (compared with the lumbopelvic pattern ($P < 0.05$) (Fig. 4) but the mean joint compression was the same in these two patterns. The free movement strategy produced a smaller joint compression than either of the coached strategies (Table 1). A similar result occurred in the end of the movement, where the free pattern resulted in the smallest joint compression with no differences between the lumbopelvic and the thoracic patterns ($P < 0.01$) (Fig. 5).

A smaller thoracic erector maximum muscle activity and a large lumbar erector maximum activity were found in the lumbopelvic pattern in comparison with the other two (Fig. 6). A similar result was observed for the muscle activity when fully upright at the end of the movement (Table 1).

4. Discussion

Overall, the findings of the study support the main hypothesis that moving the trunk from a slouch to an upright position with the lumbopelvic pattern creates less load on the lumbar spine than the thoracic pattern. The lumbopelvic strategy positioned the lumbar spine closest to the neutral posture, minimizing passive tissue stress. Neutral posture means elastic equilibrium, which is, by definition, the posture with least elastic stress and lowest joint load (Scannell and McGill, 2003).

The smaller joint compression in the free pattern might be linked with the intermediate muscle activity seen in this pattern and the unconstrained way it was executed by the participants. The lumbopelvic emphasis activates the hip flexors where psoas would impose lumbar compression. On the other hand, the thoracic pattern activated the thoracic extensors to higher levels which imposed substantial lumbar loading. Thus it appears that non-pained people have naturally found a low-stress strategy to sit upright. However a back-pained individual may find another specific strategy less painful depending on the source of pain. For

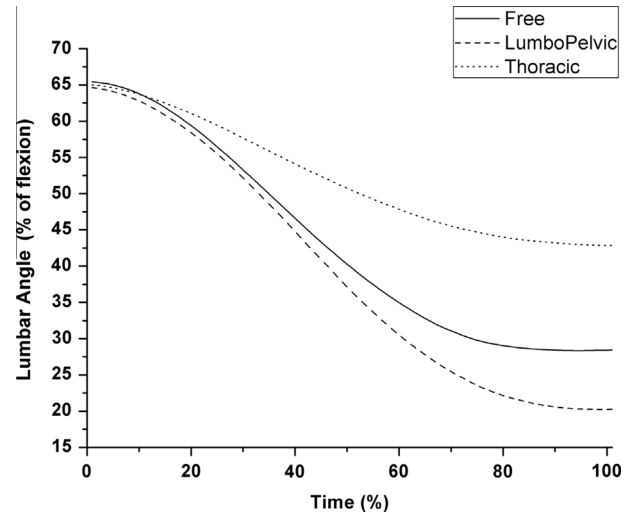


Fig. 3. Time series of mean lumbar angle (normalized by maximum angle of flexion) from slouch to upright position in the three sitting patterns across participants.

Table 1
Mean values (SD) across participants of the joint moment at the end of the movement (Final Joint Moment), mean joint compression (mean of 30% final frames), muscle activity and angles in the Free (F), Lumbopelvic (L) and Thoracic patterns (T). *P*-values with * indicate statistical significance in the post hoc comparison between patterns.

Variable	Free (F) Mean (SD)	Lumbopelvic (L) Mean (SD)	Thoracic (T) Mean (SD)	*P*-value
Final joint moment (N m)				F-L 1.00
	32.3 (3.2)	34.7 (4.7)	42.3 (5.7)	F-T 0.90
				L-T 0.60
Mean joint compression (N)				F-L 0.15
	1181 (118)[a]	1279 (112)	1367 (125)[a]	F-T 0.04*
				L-T 0.50
Max. thoracic erector activity (%)				F-L < 0.01*
	11.1 (1.2)	7.3 (1.1)	12.0 (1.5)	F-T 0.87
				L-T < 0.01*
Max. lumbar erector activity (%)				F-L 0.05*
	8.3 (0.9)	11.3 (1.3)	4.5 (0.6)	F-T < 0.01*
				L-T < 0.01*
Final thoracic erector activity (%)				F-L 0.75
	6.7 (0.9)	5.8 (1.0)	9.2 (1.3)	F-T 0.05*
				L-T < 0.01*
Final lumbar erector activity (%)				F-L < 0.01*
	2.7 (0.4)	6.2 (1.0)	2.2 (0.3)	F-T 0.46
				L-T < 0.01*
Final pelvic angle (%) (degrees)				F-L 0.01*
	55.3 (1.7)	57.3 (1.9)	50.1 (1.9)	F-T < 0.01*
	69.4 (8.4)	71.9 (10.1)	63.6 (9.7)	L-T < 0.01*
Final lumbar angle (%) (degrees)				F-L < 0.01*
	28.5 (2.6)	20.3(2.1)	42.8(3.1)	F-T < 0.01*
	19.9 (9.3)	−14.3(7.8)	−29.9 (11.5)	L-T < 0.01*

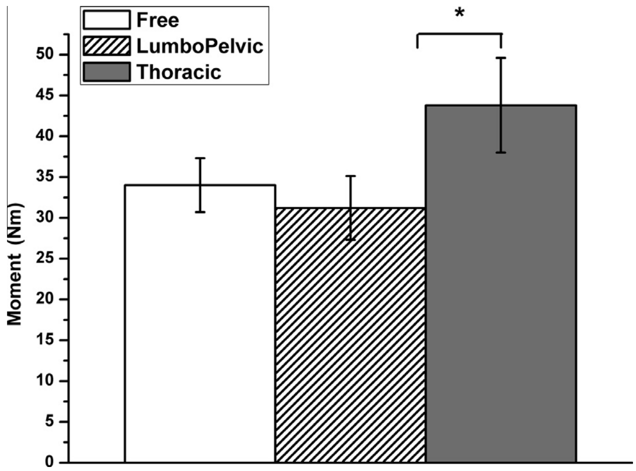


Fig. 4. Mean lumbar joint moment in the three sitting patterns across participants. Bars indicate standard deviation. The thoracic pattern generates a larger moment than the lumbopelvic pattern. *Statistically significant difference.

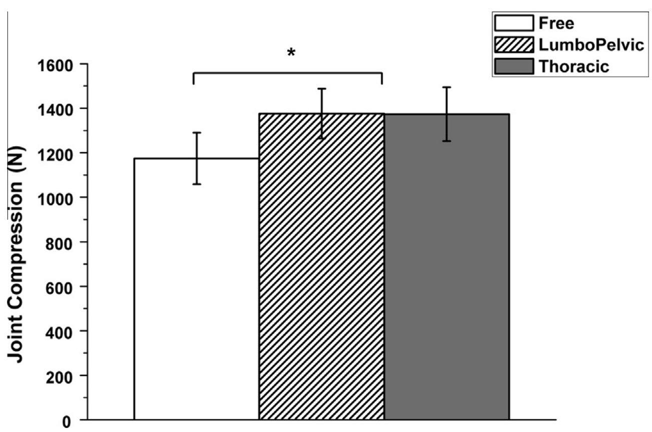


Fig. 5. Mean values of joint compression (N) at the end of the movement for the three sitting patterns across participants. Bars indicate standard deviation. The free pattern shows smaller joint compression than the others two patterns. *Statistically significant difference.

example, those back-pained individuals with spine flexion posture intolerance may benefit from a lumbopelvic sitting strategy. Dunk et al. (2009) have suggested that the majority of flexion when sitting occurs at the L5/S1 joint and it is driven principally by the rotation of the pelvis via the hips, and some people with low back and hip disorders have less ability to move their lumbopelvic region (Dankaerts et al., 2006). In this study of healthy participants we observed qualitatively differences in this motor control skill, even with no structural limitation.

The mean joint moment at the L4/L5 level during the lumbopelvic pattern was comparable with the moment produced in a four point kneeling with contralateral arm and leg extension stabilization exercise (Kavcic et al., 2004). The moment during the thoracic pattern was larger than in that particular exercise. During sitting, both right and left side back extensors are equally active. In the contralateral "birddog or quadruped" one side has a much lower activation making this a much more tolerable exercise for those with load intolerance. This comparison also highlights the relative expense in terms of spine load to sit upright in an unsupported fashion. The values for joint compression obtained during sitting were compatible with the values published by Callaghan and McGill (2001) for the same posture and are larger than the compression during standing. The muscle activity in the lumbopelvic and thoracic patterns was similar to that found by O'Sullivan et al. (2006), who postulated that a predominant activation of the thoracic erector spinae, as in the thoracic pattern, exerts higher compressive loads on the spine. The data presented here confirms this prediction.

A limitation of this data collection approach was the inability to accurately capture the passive tissue contributions throughout the complete range of motion. Passive tissue contributions throughout the range of motion are subject specific requiring extensive calibration – this was not performed. However, the highly non-linear nature of spine passive tissues would only influence joint moments at the end-range of motion, specifically when fully slouched. Therefore, the fully slouched posture would render inaccurate calculations of tissue stress and load. For this reason, the initial posture was not quantified for spine load in this study. Another limitation in this study is that the relatively small sample size, 21 subjects, might have affected the present results, particularly to explain why some statistical effects were not observed. In

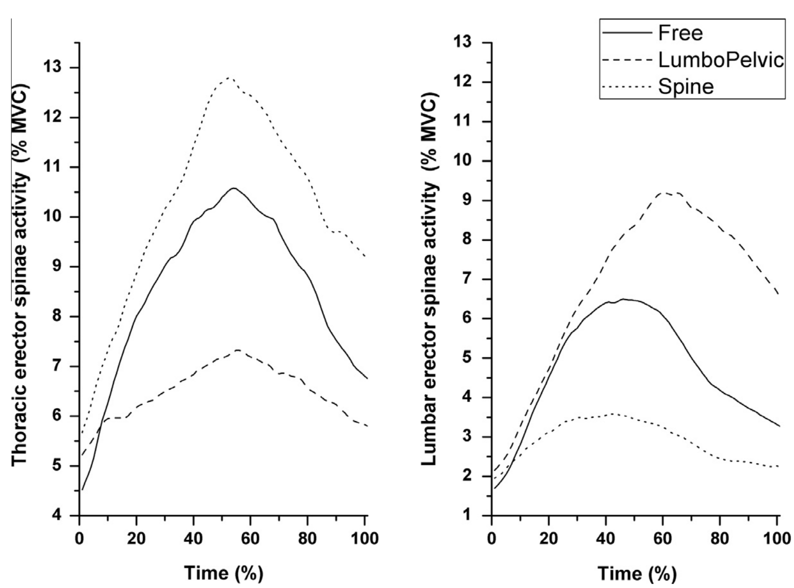


Fig. 6. Time series of mean muscle activity of the thoracic erector spinae and lumbar erector spinae (expressed as a percentage of muscular voluntary contraction – MVC) from slouch to upright position in the three sitting patterns across participants.

addition, given that we only investigated health young adults, caution should be taken to extrapolate the present results to other populations such as older populations with stiff hips.

5. Conclusion

While sitting, there are two fundamental and contrasting movement patterns to adjust the posture from a slouch position to an upright trunk position, a lumbopelvic pattern and a thoracic pattern, at least in the young and healthy participants studied here. The lumbopelvic pattern, which involves hip flexion to align the spine, generates a smaller joint moment on the lumbar spine than the thoracic pattern, which is a movement that occurs predominantly in the thoracic-lumbar spine transition region. The lumbopelvic strategy also positions the lumbar spine closest to the neutral posture minimizing passive tissue stress. This may be the strategy of choice for people with low back flexion intolerance, although the relevance of this hypothesis awaits further research.

Conflict of interest

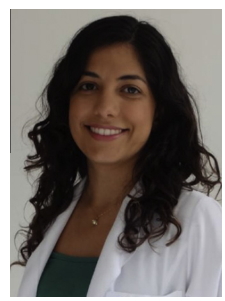
There were no conflicts of interest associated with this publication.

Acknowledgements

Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP/Brazil) for Raquel Castanharo's scholarship, Natural Sciences and Engineering Research Council (NSERC) Canada for support of scientific activity in Canada.

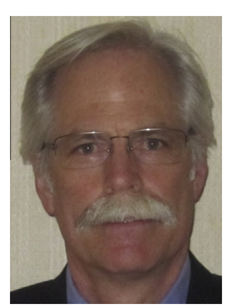
References

Brereton LC, McGill SM. Frequency response of spine extensors during rapid isometric contractions: effects of muscle length and tension. J Electromyogr Kinesiol 1998;8:227–32.
Callaghan JP, McGill SM. Low back joint loading and kinematics during standing and unsupported sitting. Ergonomics 2001;44:280–94.
Cholewicki J, McGill SM. Mechanical stability of the in vivo lumbar spine: implications for injury and chronic low back pain. Clin Biomech (Bristol, Avon) 1996;11:1–15.
Dankaerts W, O'Sullivan P, Burnett A, Straker L. Differences in sitting postures are associated with nonspecific chronic low back pain disorders when patients are subclassified. Spine (Phila Pa 1976) 2006;31:698–704.
Dunk NM, Kedgley AE, Jenkyn TR, Callaghan JP. Evidence of a pelvis-driven flexion pattern: are the joints of the lower lumbar spine fully flexed in seated postures? Clin Biomech (Bristol, Avon) 2009;24:164–8.
Endo K, Suzuki H, Nishimura H, Tanaka H, Shishido T, Yamamoto K. Sagittal lumbar and pelvic alignment in the standing and sitting positions. J Orthop Sci 2012;17:682–6.
Grenier SG, McGill SM. Quantification of lumbar stability by using 2 different abdominal activation strategies. Arch Phys Med Rehabil 2007;88:54–62.
Kavcic N, Grenier S, McGill SM. Determining the stabilizing role of individual torso muscles during rehabilitation exercises. Spine (Phila Pa 1976) 2004;29:1254–65.
McGill SM. A myoelectrically based dynamic three-dimensional model to predict loads on lumbar spine tissues during lateral bending. J Biomech 1992;25:395–414.
McGill SM. Low back disorders. 2nd ed. Evidence-based prevention and Rehabilitation. Champaign: Human Kinetics; 2007.
McGill SM, Norman RW. Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine (Phila Pa 1976) 1986;11:666–78.
McGregor AH, Hukins DW. Lower limb involvement in spinal function and low back pain. J Back Musculoskelet Rehabil 2009;22:219–22.
Mellin G. Decreased joint and spinal mobility associated with low back pain in young adults. J Spinal Disord 1990;3:238–43.
Moreside JM, Vera-Garcia FJ, McGill SM. Trunk muscle activation patterns, lumbar compressive forces, and spine stability when using the bodyblade. Phys Ther 2007;87:153–63.
Nachemson AL. Disc pressure measurements. Spine (Phila Pa 1976) 1981;6:93–7.
O'Sullivan PB, Dankaerts W, Burnett AF, Farrell GT, Jefford E, Naylor CS, et al. Effect of different upright sitting postures on spinal-pelvic curvature and trunk muscle activation in a pain-free population. Spine (Phila Pa 1976) 2006;31:E707–12.
Scannell JP, McGill SM. Lumbar posture-should it, and can it, be modified? A study of passive tissue stiffness and lumbar position during activities of daily living. Phys Ther 2003;83:907–17.
Vergara M, Page A. Relationship between comfort and back posture and mobility in sitting-posture. Appl Ergon 2002;33:1–8.
Videman T, Nurminen M, Troup JD. Volvo Award in clinical sciences. Lumbar spinal pathology in cadaveric material in relation to history of back pain, occupation, and physical loading. Spine (Phila Pa 1976) 1990;15:728–40.
Wilder DG, Pope MH, Frymoyer JW. The biomechanics of lumbar disc herniation and the effect of overload and instability. J Spinal Disord 1988;1:16–32.
Wilke HJ, Neef P, Caimi M, Hoogland T, Claes LE. New in vivo measurements of pressures in the intervertebral disc in daily life. Spine (Phila Pa 1976) 1999;24:755–62.



Raquel Castanharo is a physiotherapist and has recently completed her Master of Science degree in Biomechanics under Dr. Marcos Duarte at the University of São Paulo (Brazil). She has done an scientific internship in the Spine Biomechanics Laboratory at University of Waterloo (Canada) with the supervision of Dr. Stuart McGill.

Marcos Duarte has a BSc in Physics and MSc PhD degrees in Sciences by University of Sao Paulo, Brazil. From 1997 till 1999, he was with the Biomechanics Laboratory at The Pennsylvania State University as a post-doctoral fellow. He now is a professor of the Biomedical Engineering program at the Federal University of ABC, Brazil.His main areas of research are in the interface between Biomechanics and Motor Control of human movement.



Stuart McGill is a spinal biomechanist and Professor in the Department of Kinesiology at the University of Waterloo. He has been the author of over 300 scientific publications that address the issues of lumbar function, low back injury mechanisms, investigation of rehabilitation programs, and performance enhancement. Much of his work is summarised in his textbooks, 'Low Back Disorders: Evidence Based Prevention and Rehabilitation' and "Ultimate back fitness and performance.