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Sitting on a chair or an exercise ball: Various perspectives to guide decision making

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Abstract

Background. Prolonged sitting is recognized as a risk factor for the reporting of low back troubles. Despite the use of exercise balls in replacement of the office chair, little quantitative evidence exists to support this practice and hence motivated this research. Given the potential for several biological effects and mechanisms this study was approached with several layers of instrumentation to quantify differences in muscle activation, spine posture, spine compression and stability while sitting on an exercise ball versus a stable seat surface. Also, differences in the pressure distribution at the seat–user interface were quantified for the different seat surfaces to provide an objective perspective on the mechanism influencing perceived comfort levels.

Methods. Eight male subjects volunteered to sit for 30 min on an exercise ball and on a wooden stool. Muscle activity and spine position were used to model spine load and stability. An additional seven sat on an exercise ball and chair to examine pressure distribution over the contact area.

Findings. There was no difference in muscle activation profiles of each of the 14 muscles between sitting on the stool and ball. Calculated stability and compression values showed sitting on the ball made no difference in mean response values. The contact area of the seat–user interface was greatest on the exercise ball.

Interpretation. The results of this study suggest that prolonged sitting on a dynamic, unstable seat surface does not significantly affect the magnitudes of muscle activation, spine posture, spine loads or overall spine stability. Sitting on a ball appears to spread out the contact area possibly resulting in uncomfortable soft tissue compression perhaps explaining the reported discomfort.

Keywords: Low back; Lumbar; Gym balls; Sitting

1. Introduction

Prolonged sitting is recognized as a risk factor for the reporting of low back troubles. Sustained lumbar flexion (loss of lumbar lordosis) (Adams and Dolan, 1995) and prolonged static loading of spinal tissues (Black et al., 1996; Callaghan and McGill, 2001) have been proposed as two possible mechanisms linking sitting to back troubles. Interestingly, people who are instructed to sit in their most comfortable position over a prolonged period of time, are observed to choose a varied rather than a single comfortable position (Black et al., 1996; Callaghan and McGill, 2001). This has motivated some to declare that "dynamic" sitting with frequent posture change is beneficial and as such recommend sitting on a exercise ball. There is no shortage of on-line marketers who recommend replacing an office chair with one of their exercise balls claiming to reduce pain, and improve spine posture and balance, to name just a few. Despite the prevalent use of exercise balls in replacement of the office chair, little quantitative evidence exists to support these claims and hence forms the purpose of this paper.

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Although there are few data available on spine motion and motor patterns while sitting on an exercise ball, there has been some effort dedicated to assessing the comfort levels of different office chairs (e.g., Vergara and Page, 2000; Bendix et al., 1985; Van Dieen et al., 2001). In a recent study on exercise balls, Gregory et al. (in press) observed that people found sitting on the balls less comfortable over time than an office chair while performing office tasks. Although, the most direct method to assess seat comfort is with subjective ratings (Richards, 1980), pressure distribution has been identified as an objective method of assessment with a strong association to subjective comfort ratings (Yun et al., 1992; Kamijo et al., 1982; de Looze et al., 2003).

The purpose of this paper was to assess the possible effects of exercise balls with several layers of instrumentation. Specifically, differences in muscle activation, spine posture, spine compression and stability were quantified while sitting on an exercise ball versus a stable seat surface. As well, differences in the pressure distribution at the seat–user interface were quantified for the different seat surfaces to provide an objective perspective on a possible mechanism influencing perceived comfort levels.

2. Methods

Two separate studies were performed. The first (Study I) assessed torso muscle activation, spine load and stability in men sitting on an exercise ball and on a stool. The second study (Study 2), assessed the pressure distribution over the buttocks and posterior thighs while men sat on a stool, an office chair and an exercise ball. Two additional conditions involving the office chair were assessed—one using the back rest and arm rests and the other not. Given the difference in paradigms they will be described separately.

2.1. Study 1: Torso muscle activity, spine load and stability

Eight male subjects [mean age of 24 years (SD = 4), height of 180.3 cm (SD = 7.7) and weight of 83.9 kg (SD = 12.4)] volunteered to participate in this study. Subjects had no history of low back pain. Prior to testing, subjects' age, height, weight, and breadth dimensions at the feet, ankles, knees, hips, hands, wrists, elbows, and shoulder were obtained while standing in the anatomical position. They sat for 30 min on an exercise ball and 30 min on a wooden stool while electromyography and three-dimensional lumbar position were measured every 5 min. These data were input into a series of biomechanical models in order to calculate a measure of L4–L5 compression and spine stability. All procedures were approved by the University of Waterloo Office for Research Ethics.

2.1.1. Data collection

Both seat surfaces (an inflatable exercise ball and a wooden stool (see Fig. 1A and B)) did not have a back rest in order to isolate the comparison specifically to the effects of the ball or chair supporting surface. The wooden stool stood 45 cm high and had a round seat surface with a diameter of 33.5 cm. This resulted in a knee angle of 90° in our cohort of men who did not have a wide variance in their height. The ball was inflated to ensure similar angles. The seating assignment presentation order was randomized. Subjects sat on each surface for a total of 30 min while watching a movie. The only instructions provided to the subjects prior to the start of their 30-min sit, was to maintain an upright sitting posture. These instructions were not reinforced throughout the sit duration but no subject sat with their elbows on their knees for example. While subjects were sitting, measures of muscle surface EMG and three-dimensional spine posture were collected every 5 min for 5 s in dura-

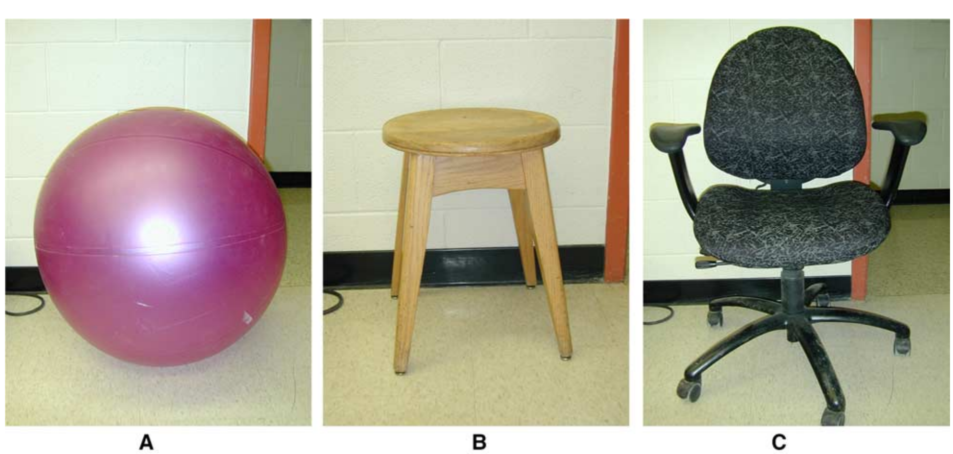

Fig. 1. Three different seat surfaces. An exercise ball (A), a wooden stool (B) and a padded office chair (C). A and B were used in the muscle and joint analysis (Study 1), and all three seat surfaces were used in the pressure distribution analysis (Study 2).

tion. In between the 30-min sit trials, subjects were required to stand and walk around the lab area for 10 min to provide a rest/change phase of muscle activation and low back loading (after Callaghan and McGill, 2001).

2.1.2. Instrumentation

2.1.2.1. Electromyography. Fourteen channels of electromyography (EMG) were collected from electrodes placed over the following muscles bilaterally; rectus abdominis, oblique internus, oblique externus, latissimus dorsi, thoracic erector spinae (longissimus thoracis and iliocostalis at T9), lumbar erector spinae (longissimus and iliocostalis at L3) and lower lumbar erector spinae (1 cm lateral to L5). The skin was shaved and cleansed with a 50/50 H_2O and ethanol solution. Ag–AgCl surface electrodes were positioned with an inter-electrode distance of about 3 cm. The EMG signals were amplified and then A/D converted with a 12-bit, 16-channel analog to digital (A/D) converter at 1024 Hz. Each subject was required to perform a maximal contraction of each measured muscle for normalization of each channel. For the abdominal muscles each subject, while in a sit up position and manually braced by a research assistant, produced a maximal isometric flexor moment followed sequentially by a right and left lateral bend moment and then a right and left twist moment (note: little motion took place). For the extensor muscles, a resisted maximum extension in the Biering–Sorensen position was performed (McGill, 2002). The EMG signal was normalized to these maximal contractions, full wave rectified and low-pass filtered with a second order Butterworth filter. A cut-off frequency of 2.5 Hz was used to mimic the frequency response of the torso muscles (Brereton and McGill, 1998).

2.1.2.2. Three-dimensional positioning of the lumbar spine. Lumbar spine position was measured about three orthogonal axes using a 3 Space IsoTRAK electromagnetic tracking instrument (Polhemus Inc., Colchester, VT, USA). This instrument consisted of a single transmitter that was strapped to the pelvis over the sacrum and a receiver strapped across the ribcage, over the T12 spinous process. Thus, the position of the ribcage relative to the sacrum was measured (lumbar motion).

2.1.3. Data processing

2.1.3.1. Calculating a stability index. The analysis of stability (Euler stability) was performed using a method documented by Cholewicki and McGill (1996) and involves several interdependant models. For the interested reader, these models are described in detail by Cholewicki and McGill (1996), however a brief description is provided here (refer to Fig. 2A and B for a flow chart of the cascading steps involved in the stability analysis). The first model is an eight-segment link segment model that uses external force measures, joint kinematics and segment anthropometrics to calculate reaction forces and moments acting at the L4–L5 intervertebral joint. The L4–L5 moments are used to ultimately drive the optimization routine that determines the muscle force profiles. The reaction forces are used in determining the shear and compression forces at the L4–L5 joint. The second model is the "Lumbar Spine model" that consists of an anatomically detailed, three-dimensional ribcage, pelvis/sacrum and five intervening vertebrae. Over 100 laminae of muscle are included together with and the passive tissues which are represented as torsional, lumped parameter stiffness elements are modeled about each axis. This model uses the measured 3D spine motion data which is partitioned to the appropriate rotation for each of the lumbar vertebral segments (after White and Panjabi, 1978). Muscle lengths and velocities are determined from their motion and attachment points on the dynamic skeleton of which moves according to the measured lumbar kinematics obtained from the subject. As well, the orientations of the vertebral segments along with the stress/strain relationships of the passive tissues were used to calculate the restorative moment created by the spinal ligaments and discs. The third model, termed the "distribution–moment model" (Guccione et al., 1998; Ma and Zahalak, 1985), is used to calculate the muscle force and stiffness profiles for each muscle. The model uses the normalized EMG profile of each muscle along with the calculated values of muscle length and velocity of contraction to calculate the active muscle force and any contribution from the passive elastic components. When input to the spine model, these muscle forces are used to calculate a moment for each of the three axes of the six intervertebral joints. An optimization routine assigns an individual gain value to each muscle force (estimated from EMG, physiological cross-sectional area, instantaneous length and velocity) in order to create a total moment about the intervertebral joint that matches those calculated by the link segment model to achieve mathematical validity. The objective function for the optimization routine operates to match the moments with a minimal amount of change to the EMG driven force profiles. The adjusted muscle force and stiffness profiles are then used in the calculations of L4–L5 compression and shear, as well as spine stability.

In the final step the stability index was obtained by calculating a level of potential energy in the spinal structure for each of the 18 degrees of freedom (three rotational axes at six lumbar joints) resulting from the combined potential energy existing in both the active and passive spinal structures, minus any work done from external loads. The 18 values of potential energy were formed into an 18×18 Hessian Matrix and diagonalized. The determinant of this matrix represented an index of spine stability. While the lowest eigenvalue indi-

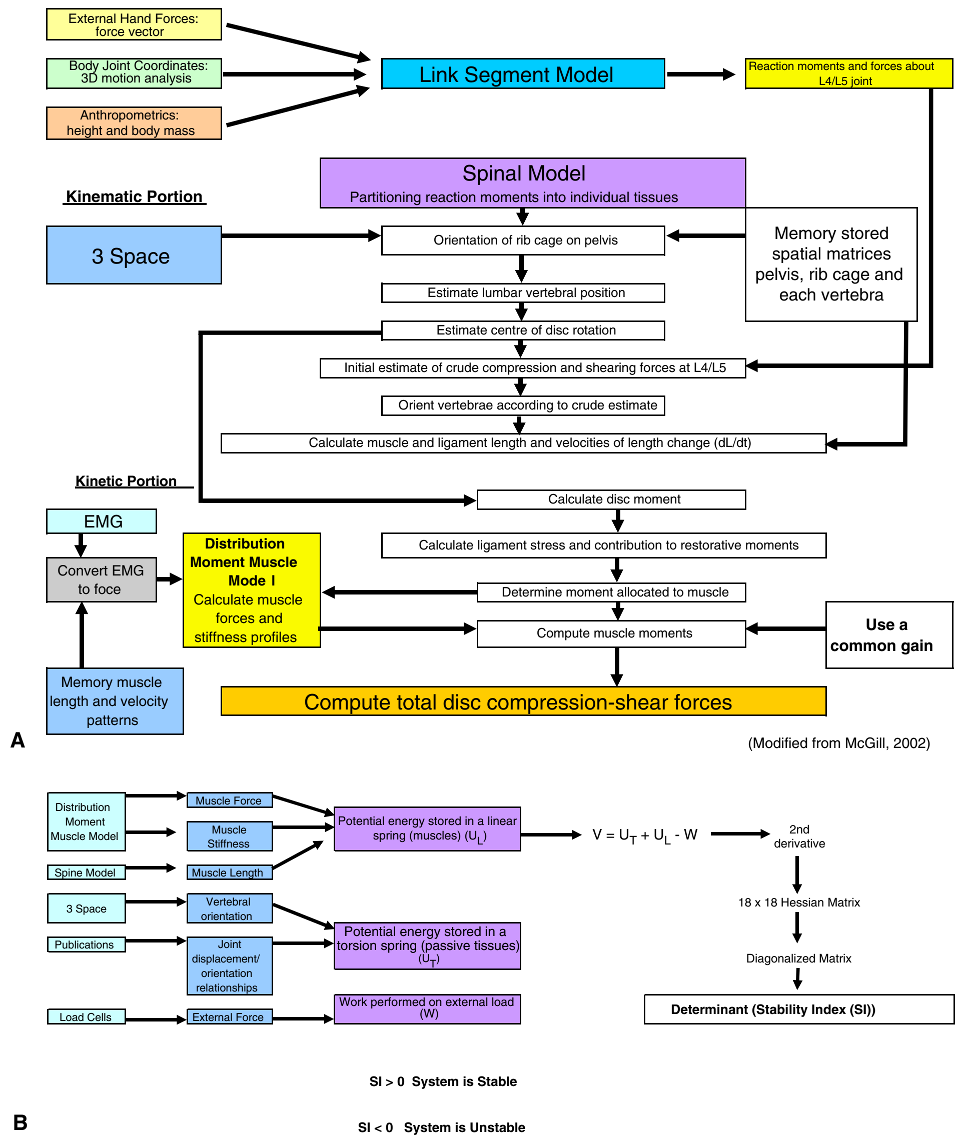

Fig. 2. Flow Charts of Models. Kinetic and kinematic data are input to the trunk segment model which generates estimates of joint moment and force. The anatomically detailed model driven by biological signals uses these variables as input together with muscle force and stiffness data from the distribution–moment (D–M) model. (A) illustrates the cascade of models and processes to obtain the stability index.

cates the mode of potential buckling as the critical value is approached, we simply needed an overall indication of relative stability and, in this way, selected the determinant as the index.

2.1.3.2. Statistical analysis. A two-way, repeated measures analysis of variance was conducted with time and seat surface as the independent variables. The interaction term was assessed to determine if the patterns of stability, compression, muscle activation profiles and spine flexion over time were significantly different between the two seat surfaces. A Bonferoni correction of the significance level, which takes into account the number of different conditions, was used, and a p-value of 0.004 or less was considered significant for all tests.

2.2. Study 2: Pressure distribution analysis

Seven male subjects, different from those studied above, sat on an exercise ball, a wooden stool and a padded office chair (POC) (see Fig. 1A–C, respectively). Two conditions on the office chair included the subjects sitting without using either the back or arm rests and repeating the protocol while using both (see Fig. 3).

2.2.1. Data collection

2.2.1.1. Sitting trials. While the ball and stool were previously described, the office chair height was adjusted to ensure that the subjects' knees were bent to 90° in the seated posture. While sitting on the office chair, subjects were instructed to sit upright without using either the

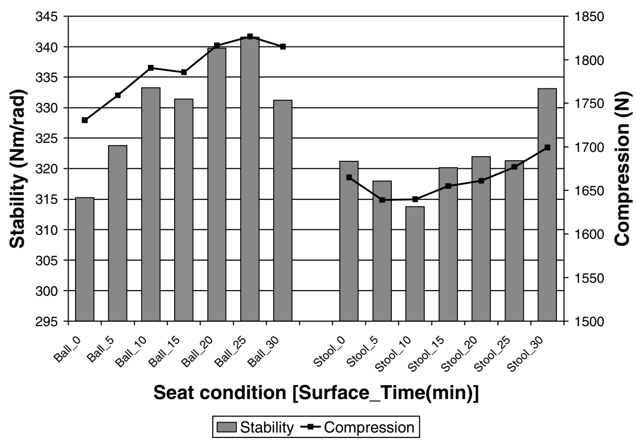

Fig. 3. Plot of spine stability and L4–L5 compression measured every 5 min for a total duration of 30 min on the ball and on the stool seat surface. While it appears that the pattern of both stability and compression over the 30-min sit are different between the two seat surfaces, the increased variability across subjects prevents a significant interaction effect between seat surface and time.

back rest or arm rests and then to lean back into the back rest and use the arm rests. Sitting bouts on the three seat surfaces were randomized. Subjects repeatedly sat on each surface for a total of 5 s, and this was repeated three times. Pressure measurements were obtained over the 5 s. In between each of the trials subjects were required to stand and then reposition themselves on the seated surface.

2.2.2. Instrumentation

2.2.2.1. Pressure measurement system. The seat–user interface pressure distribution was measured using a pressure mapping system (I-Scan, Tekscan Incorporated, Boston, USA). The sensor mat used was an ultra-thin (0.004″, 0.10 mm) flexible printed circuit with 2016 individual sensing elements or cells organized in a 42×48 array. Before the study, the pressure mat was calibrated up to 1379 kPa (200 PSI—note that the pressure output contour figures have the units of PSI) using a uniform pressure applicator. During the collection, the pressure mat was placed only on the specific seat surface.

2.2.3. Data processing

2.2.3.1. Quantifying differences in pressure distribution. Four different variables were quantified from the pressure array of data. The first variable 'Total Force' was the total force measured within the sensor area. 'Contact Area' was the area encompassed by the edge of the loaded sensors. 'Peak Contact Pressure' was the force in the highest pressure area. The last analysis was a frequency count of all of the sensors on the mat recording a specific range of pressures. This analysis was used to provide a measure of variety of the pressure distribution across the user–seat interface. The range of pressures measured were <0.05 PSI (this represents the area of no contact on the pressure mat), 0.06–1, 1.1–2, 2.1–3, 3.1–4, 4.1–5, 5.1–10, 10.1–20, 20.1–30, >30 PSI. (The units of PSI are used here since these are the units used by the system for pressure contour mapping. Convert PSI into kPa by multiplying by 6.895.) All of these measures were taken at the instant of maximum contact area over the 5 s collection.

2.2.3.2. Statistical data analysis. A one-way, repeated measures ANOVA was used to identify any significant differences in the pressure distribution output measures across seat conditions. A Tukey's post hoc analysis was performed to identify any specific differences between seat conditions.

3. Results

3.1. Study 1: Torso muscle, spine load and stability

There was no significant interaction between time and seat conditions for any of the 14 muscles. The activation values remained quite low, for example, the abdominal muscles (rectus abdominis, external and internal obliques) amplitudes ranged from 1% of maximum voluntary contraction (MVC) to 2.8%MVC. Back muscle amplitudes ranged from 1.3%MVC to 4.8%MVC. The highest amplitudes were measured in the upper erectors and the lower lumbar muscles. No significant difference in spine flexion angle was observed.

Calculated stability and compression values appear to be higher for the ball condition over the 30-min sit duration (Fig. 3). However, there was no significant interaction observed between seating surface and time.

3.2. Study 2: Pressure distribution analysis

Four measures were obtained to quantify differences in the pressure distribution across the four different seat surfaces (Fig. 4). There was no difference in 'Total force'.

The contact area of the seat–user interface (Fig. 5) was greatest on the exercise ball (793 cm^2) and was significantly greater that that on each of the other seat surfaces ($p < 0.01$). The stool produced the smallest contact area out of all the seat surfaces tested (488 cm^2). When sitting on an office chair, putting pressure on the back rest reduces the contact area at the seat pan (628 cm^2 with the back rest and 664 cm^2 without the back rest).

The average peak contact pressure recorded on the stool was 158 kPa (23 PSI) (Fig. 6). This value was higher that that measured on the three other support surfaces. Among the ball and office chair, no significant difference existed. As well, using the back rest did not significantly affect the peak contact pressure.

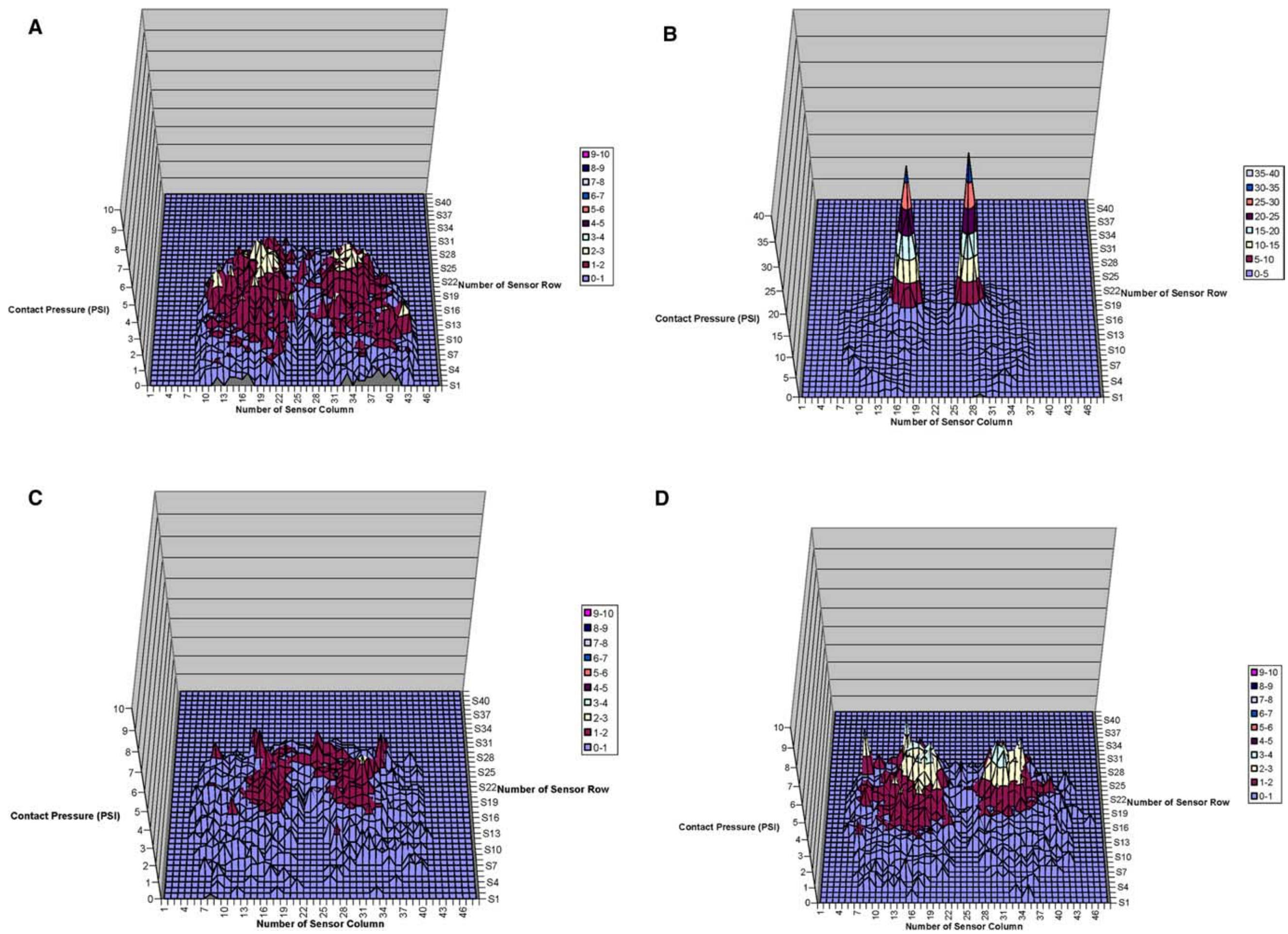

Fig. 4. Example of output pressure distribution while sitting on the exercise ball (A), wooden stool (B), POC while using the back rest (C) and POC while not using the back rest (D). For example, the pressure concentrations under each ischial tuberosity is clear on the stool while the distributed soft tissue pressure over the buttocks and posterior thighs is clear while sitting on the ball.

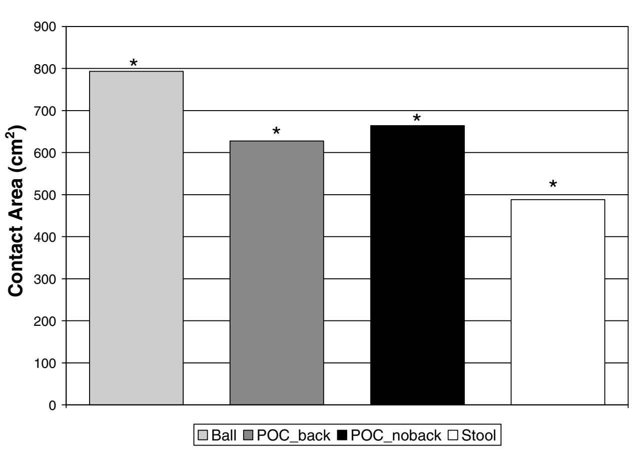

Fig. 5. Recorded measures of contact area on each of the four seat surfaces (POC_back = padded office chair using the back and arm rests, POC_noback = padded office chair not using the back and arm rests). * indicates that the contact area while sitting on the particular seat surface is significantly different than the contact area for all the other seat surfaces ($p < 0.01$). Sitting on the exercise ball results in the greatest contact area and sitting on the stool produces the smallest contact area.

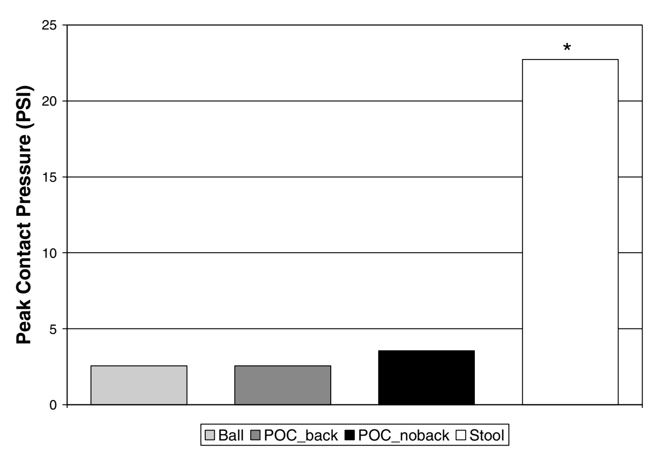

Fig. 6. Recorded measures of peak contact pressure on each of the four seat surfaces (POC_back = padded office chair using the back and arm rests, POC_noback = padded office chair not using the back and arm rests). * indicates that the peak contact pressure while sitting on the particular seat surface is significantly different then the peak contact pressure for all the other seat surfaces. Sitting on the stool results in a significantly higher peak contact pressure then that produced on the three other seat surfaces ($p < 0.01$).

The last analysis assessed variability in the pressure distribution while sitting on each of the four support surfaces to assess stress gradients (seen in Fig. 4 in the contour surface). The range and distribution of pressures was greatest for the stool condition as indicated by a significant difference ($p < 0.01$) in the number of sensor cells in a specific pressure range, together with the fact that much higher amplitudes were uniquely observed such that there were no comparative values observed in the other sitting conditions.

4. Discussion

The results of this study suggest that prolonged sitting on a dynamic, unstable seat surface does not significantly affect the magnitudes of muscle activation, spine posture, spine loads or overall spine stability. Thus, this study failed to find any beneficial changes to either spine stability or compression while sitting on the ball over 30 min. In contrast, sitting on a ball spreads out the contact area into tissues not usually loaded during sitting possibly resulting in uncomfortable soft tissue compression—the candidates being the gluteal and hamstring muscles.

According to previous research there are patterns of pressure distribution across the seat–user interface which are highly associated with perceived ratings of comfort. Further, in a meta-analysis de Looze et al. (2003) concluded that pressure distribution provided the clearest association with ratings of comfort over such other measures as posture and movement, electromyography, spinal load and foot swelling. Kamijo et al. (1982) reported that comfortable car seats are characterized by mean pressure levels of 5.79 kPa under the ischial region and 2.89 elsewhere. Both Kamijo et al. (1982) and Yun et al. (1992) found that uniformity in pressure distribution is associated with local discomfort. When interpreting the results of the current study, it is obvious that although the stool creates a high variety in the pressure magnitudes across the contact area, which is beneficial for level of comfort, the peak pressure was extremely high. This probably contributes to reports of discomfort while sitting on such hard surfaces. The larger contact area on the ball could potentially act as a cause of discomfort given that there is a greater amount of soft-tissue compression compared to sitting on the office chair. This compression may lead to circulation blockage acting as a mechanism of pain, soreness and numbness (de Looze et al., 2003).

Despite the static analysis used, a lack of significant changes in EMG from sitting on a dynamic versus static seat surface was supported in previous work by Van Dieen et al. (2001). These researchers failed to find a difference between erector muscle EMG and spine posture changes in the sagittal plane with static office chairs versus dynamic office chairs where the seat and back rest were movable. They did report that sitting on a dynamic chair reduced spine compression as measured by a gain in standing stature. Leivseth and Drerup (1997) reported that when sitting in an office chair, using the back rest also reduces the compressive loading on the spine compared to an upright sitting posture. In the study by Van Dieen et al., both the static and dynamic office chairs had back rests, so the results on compression could be isolated to the effect of the chair type. From these studies it appears that both the amount of spine motion and the use of a back rest are two factors that can act to reduce the risk during prolonged sitting. Potentially, the insignificant difference in compression and stability observed between the exercise ball and wooden stool is due to the fact that on both seat surfaces the lumbar spine was unsupported thereby requiring a prolonged low-level activation of the spinal musculature. This continual low-level activation may have prevented relaxation of the Type I muscle fibres even though changes in the EMG patterns were occurring as a result of the dynamic movements (as suggested by Van Dieen et al., 1993). As well, the level of liability on the exercise ball may have been insufficient enough to cause increased co-contraction patterns over those that were already present to support the upright sitting spine posture. This is in contrast to the observations we made several years ago where performing some abdominal exercises on a ball greatly increased co-contraction when compared to a stable surface (Vera-Garcia et al., 2000). As well, the spacing between the subjects' feet in the frontal plane affects the base of support and could act to reduce the instability created while on the ball. Initial foot placement was set to shoulder width for each subject, however, foot movement over time was not controlled, possibly contributing to the variability observed across subjects.

Several limitations exist due to the protocol used. This was not a fully dynamic assessment of the differences in sitting on a stable versus unstable surface. The analysis only considered a 5-s sample of posture and EMG patterns taken every 5 min. Collection over the entire 30 min would enable more complete understanding of any difference between the dynamics of the two seat surfaces. Further, perhaps there may be changes in sitting dynamics for durations longer than 30 min. Another consideration is that the ball and stool affect deeper muscles than the superficial muscles measured here. However, the smaller local muscles have a smaller effect on stability and compression (Kavcic et al., 2004).

It appears that sitting on an exercise ball has little effect on spine loads, muscle activity and the resulting spine stability at least in the type of static sitting task studied here. Perhaps sitting on an exercise ball during more dynamic tasks (for example reaching for a phone) would affect muscle activation. However, sitting on an exercise ball produces increased soft tissue compression possibly explaining the increased subjective feelings of discomfort over time noted by other researchers.

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