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PREVENTION & REHABILITATION: ORIGINAL BIOMECHANICAL RESEARCH AND HYPOTHETICAL MODEL

Lumbopelvic muscle activation patterns in three stances under graded loading conditions: Proposing a tensegrity model for load transfer through the sacroiliac joints

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KEYWORDS
Sacroiliac joint;
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Summary *Purpose:* According to the conventional arch model of the pelvis, stability of the sacroiliac joints may require a predominance of form and force closure mechanisms: the greater the vertical shear force at the sacroiliac joints, the greater the reliance on self-bracing by horizontally or obliquely oriented muscles (such as the internal oblique). But what happens to the arch model when a person stands on one leg? In such cases, the pelvis no longer has imposts, leaving both the arch, and the arch model theory, without support. Do lumbopelvic muscle activation patterns in one-legged stances under load suggest compatibility with a different model? This study compares lumbopelvic muscle activation patterns in two-legged and one-legged stances in response to four levels of graded trunk loading in order to further

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our understanding the stabilization of the sacroiliac joints.
Methods: Thirty male subjects experienced four levels of trunk loading (0%, 5%, 10% and 15% of body weight) by holding a bucket at one side, at three conditions: 1) two-legged standing with the bucket in the dominant hand, 2) ipsilateral loading: one-legged standing with the bucket in the dominant hand while using the same-side leg, and 3) contralateral loading: one-legged standing using the same leg used in condition 2, but with the bucket in the non-dominant hand. During these tasks, EMG signals from eight lumbopelvic muscles were collected. ANOVA with repeated design was performed on normalized EMG's to test the main effect of load and condition, and interaction effects of load by condition.
Results: Latissimus dorsi and erector spinae muscles showed an antagonistic pattern of activity toward the direction of load which may suggest these muscles as lateral trunk stabilizers. Internal oblique muscles showed a co-activation pattern with increasing task demand, which may function to increase lumbopelvic stability ($P < 0.05$). No unilateral pattern of the internal obliques was observed during all trials..
Conclusions: Our results suggest that the lumbopelvic region uses a similar strategy for load transfer in both double and single leg support positions which is not compatible with the arch analogy. Our findings are more consistent with a suspensory system (wire-spoke wheel model). If our proposed model holds true, the pelvic ring can only be integrated by adjusting tension in the spokes and by preserving rim integrity or continuity. Thus, we propose that in order to restore tension integrity throughout the pelvic ring, efforts to unlock restrictions, muscular correction of positional faults and lumbopelvic or even respiratory exercises following sacro-iliac joint dysfunctions must be taken into consideration. Our hypothetical model may initiate thinking and act as a guide to future work based on a biomechanical approach to the problem of sacroiliac joint dysfunction..

Introduction

The aim of this study was to compare the lumbopelvic muscle activation patterns between two-legged standing and one-legged standing positions in response to four levels of trunk loading in order to further our understanding of the stabilization of the sacroiliac (SI) joints.

Low back pain (LBP) affects a remarkable percentage of the human population every year. A systematic review of epidemiological studies on LBP found that the point prevalence of LBP was as high as 33%; one year prevalence was as high as 65% and lifetime prevalence was as high as 84% (Walker, 2000). These results indicate that LBP is widespread and deserves to be studied in depth.

The lumbopelvic region and especially the SI joints, in regard to their intermediary position, play an important role in transferring loads generated by upper body weight and gravity while sitting, standing and walking. It is suggested that SI dysfunction can contribute to LBP (Schwarzer et al., 1995). Therefore, understanding the stability mechanisms in the SI joints is important from both a diagnostic and a therapeutic standpoint.

According to the conventional arch model, the pelvis is structured as a Roman arch with the sacrum as the keystone, wedged between the two iliac bones (form closure). See Fig. 1

Two-legged standing supports the arch analogy. The legs can be seen as the imposts of an arch, allowing the sacrum to transfer forces to the articular surfaces of the SI joints through compression, which can prevent unnecessary muscle forces, and most studies on the stability of SI joints are in general agreement with the arch theory. In this model, the main source of loading is the sacrum, taking load from the spine and transferring it to the hips through

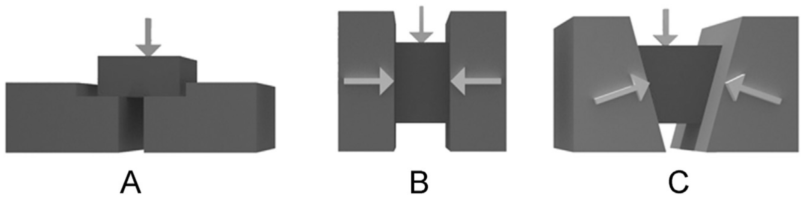

Figure 1 (A) Form closure: the object remains in place, independent of the exerted load. (B) Force closure: the object can only remain in place when continuous additional transversely oriented forces are applied to resist movement by friction. (C) Combination of form and force closures.
Adapted from Snijders et al. (1998).

compression forces transmitted along the ilium into the acetabulum (Vleeming et al., 1990a,b). The anatomical orientation of the SI joints creates form and force closure mechanisms, as described by Snijders et al. (Snijders et al., 1993, 1998). Pel et al. (2008) performed a biomechanical analysis of SI joint stability in upright standing posture using a validated static 3-D simulation model. They concluded that TrA and pelvic floor muscles stabilize the position of the sacrum between the iliac bones through a self-bracing mechanism (the pelvic arch).

Nonetheless, absolute consensus has not yet been reached and the arch theory has not been evaluated in single-legged standing. Indeed, when a biped stands on a single leg, the whole concept of an arch may fall apart. One-legged standing creates an arch with one of the imposts missing; a true Roman arch would not be able to stand under such circumstances, yet bipeds do it all the time. This suggests that the arch theory, much like the one-impost arch, lacks essential support.

Snijders et al. (1998) found a decrease of IO activity (contralateral to support side) in one-legged resting support (bus-stop position), and theorized the cause of the decrease of IO activity to be *decrease of gravitational load*. They concluded that this observation supports the biomechanical model of self-bracing of the SI joints. They assumed that in one-legged standing, shearing loads will increase on the ipsilateral SI joint and decrease on the contralateral SI joint. Hungerford et al. (2003) used standing hip flexion tasks to determine whether muscle activation of the supporting leg in one-legged standing was different between control subjects and subjects with SI joint pain. Although they used a single leg position in their study, their work focused on evaluating temporal activity (onset of EMG activity) which does not address the issue of load transfer through the SI joints. In addition, EMG data were recorded from the seven trunk and hip muscles on the ipsilateral side in one-legged standing. In fact, they never measured what was happening on the contralateral side.

In contrast to the pelvic arch theory, some anatomists (Kapandji et al., 1982; Grant, 1989) and some clinicians (Dijkstra, 2007; DonTigny, 2007) theorize that the sacrum is hanging from the iliac bones. Levin (2007) proposes a suspensory system for the stability of the SI joints, a "wire-spoke bicycle wheel" model. In a bicycle wheel, tension-loaded spokes transmit compressive loads from the frame and the ground. The hub remains suspended in its tension network, and the compression loads distribute around the rim. The compression elements are discontinuous and behave in a counterintuitive way. Rather than becoming the primary support elements of the system, as they would be in a pillar or wagon wheel model, the compression elements become secondary to the tension support network. According to this model, the sacrum is suspended in its network much as the "hub" of the bicycle wheel is suspended in its tensioned spokes. However, might the actual situation in the body be different because the pelvic ring, in contrast to a bicycle wheel, is not a closed ring and is open at the top and also articulated at symphysis pubis which makes the pelvic rim susceptible to opening?

In light of the aforementioned theories we ask: In the pelvic arch model, how can stability of the SI joints be achieved during one-legged standing, in the absence of one of the imposts? What muscle activation patterns manifest in one-legged standing? To attain stability of the SI joints in one-legged standing, does the lumbopelvic region use muscle activation patterns similar to those used in double-legged standing? We theorized that higher levels of task demand (compared to standard standing position) may be required to effectively provoke the behaviour of the lumbopelvic stabilization system in the process of load transfer through SI joints. To solicit muscle activity patterns in two-legged and one-legged standing positions, we used increasing loads in addition to superincumbent weight to challenge the SI joints.

Materials and methods

Study design

This study was a two-factor within subjects design in order to compare muscle activation patterns under different graded loading conditions. The lumbopelvic muscle activation levels, as the dependent variable, were assessed in 4 levels of loading (load factor) and in three different loading conditions (condition factor), as independent variables.

Participants

Thirty male subjects with no history of back pain or surgery (Mean (SD) age 24.7 (4.1) year, height 172 (4.4) cm, and weight 68 (8.0) kg) were recruited from the university population. Participants completed an informed consent form approved by the University Research Ethics Committee.

Experimental setup

Prior to testing, each subject's height and weight were taken. Based on the pilot study findings, Zero, 5, 10 and 15 percent of subjects' body weight (BW) was respectively calculated to determine 4 levels of loading. For EMG measurements, the areas for electrode placement were marked, shaved, grated and finally cleaned with an alcohol swab. Surface, reusable and bipolar EMG electrodes (Biometrics Ltd, UK) with 10 mm diameter and 20 mm centre to centre inter-electrode distance was covered with a solid conducting gel and then attached to 4 pairs of lumbopelvic muscles and secured with adhesive tape. Electrode positions were based on previous anatomical investigations (Ng et al., 1998; Hermens et al., 2000). The electrodes for latissimus dorsi (LD) were placed over the muscle belly at T12 level and along a line connecting the most superior point of the posterior axillary fold and the S2 spinous process. For the IOs, electrodes were placed 1 cm medial to the anterior superior iliac spine (ASIS) and beneath the line joining both ASISs. For the erector spinae (ES), electrodes were placed 3 cm lateral to L3 spinous process. For the biceps femoris (BF), electrodes were placed midway between the ischial tuberosity and the caput fibulae. The ground reference electrode was placed around the wrist by a strap. We decided to select muscles that are located in the "sling

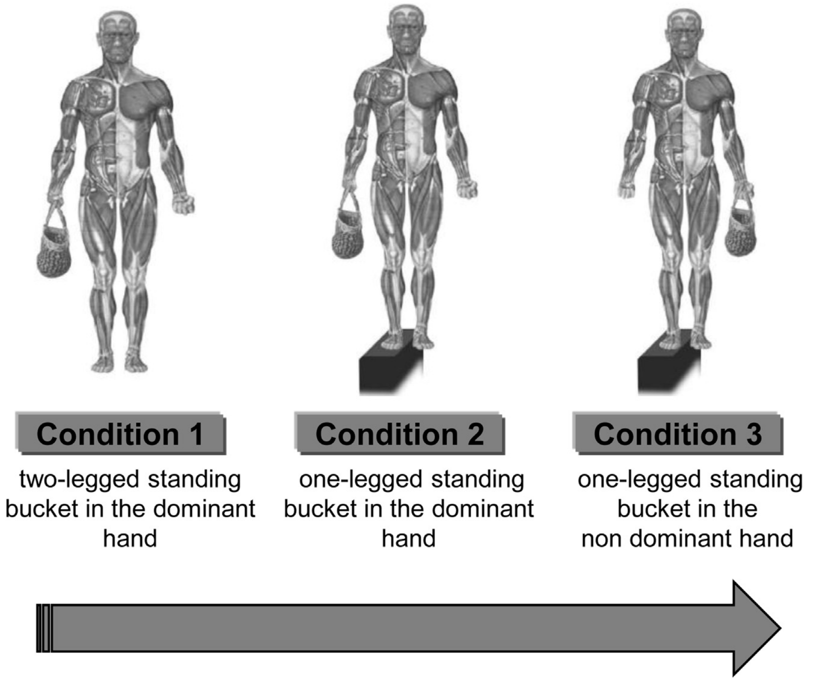

Figure 2 Subjects positioned in erect standing posture in front of a checkered mirror and held single handed load under three support conditions (1. two-legged standing with the bucket in the dominant hand, 2. ipsilateral loading: one-legged standing with the bucket in the dominant hand while using the same-side leg, and 3. contralateral loading: one-legged standing using the same leg used in condition 2, but with the bucket in the non-dominant hand). For trunk loading, each subject held a bucket of pre-determined load levels (0, 5, 10 and 15%BW) in his hand. The subject depicted is assumed to be right hand dominant.

system" described in the previous studies, which are supposed to have a role in pelvic self-bracing mechanisms. Since we utilized surface electrodes, crosstalk from surrounding muscles may have been possible, specifically in the case of IO muscle. Although we placed the electrodes on sites of less coverage by the TrA muscle we might also record TrA activity. Because we purposely decided to record activity from the transversely oriented abdominal muscles, differentiation between their activities was not necessary. Studies with more EMG channels and fine wire electrodes may be helpful in the future.

Procedure

Each subject was positioned in erect standing posture, with the feet one foot (15 cms) apart, in front of a checkered mirror, and was asked to look in a horizontal direction at the checkered mirror. During all trials, subjects were asked to maintain a level pelvis and postural alignment. Visual feedback together with verbal encouragement was used to ensure pelvic positioning.

For trunk loading, each subject held a bucket at his side under four predetermined load grade (load factor) using

Table 1 Results of two-factor within subject analysis of variance.

Muscles	Load		Condition		Load * condition	
	(F)	*(P)*	*(F)*	*(P)*	*(F)*	*(P)*
S. biceps femoris	25.51	0.00	1.91	0.15	8.75	0.00
N. biceps femoris	10.76	0.00	17.06	0.00	12.73	0.00
S. erector spinae	11.29	0.00	51.85	0.00	40.69	0.00
N. erector spinae	33.97	0.00	34.56	0.00	21.11	0.00
S. latissimus dorsi	37.21	0.00	27.08	0.00	28.12	0.00
N. latissimus dorsi	100.25	0.00	40.59	0.00	39.24	0.00
S. internal oblique	14.60	0.00	5.35	0.00	9.27	0.00
N. internal oblique	12.94	0.00	9.10	0.00	10.22	0.00

F (F-statistics), *P* (significance), *N* (Non-support side), *S* (Support side).

three stances (support condition) presented in a fixed order (Fig. 2). A problem with within-subjects designs is that they are subject to carry-over effects or order effects. To counter this, the levels of loading for each stance were arranged in a randomized order (random assignment). We divided our subjects into three groups (between-subject factor) and each group performed the graded loading tasks in a different order. If we define each level of trunk loading, 5% of BW, 10% of BW and 15% of BW respectively by A, B and C level, our three different orders of loading levels were: ABC, BCA and CAB. On a step with riser height of seven inches, three stances (support conditions) were used: 1) subjects held a bucket at their side in their dominant hand while standing with their body weight balanced between both legs (two-legged standing; first condition); 2) subjects held a bucket at their side in their dominant hand while standing with their body weight balanced on their ipsilateral leg (one-legged standing - ipsilateral loading; second condition); 3) subjects held a bucket at their side in their non-dominant hand while standing with their body weight balanced on the leg used in stance 2 (one-legged standing - contralateral loading; third condition). Ten seconds of EMG data were collected at a sampling rate of 1000 Hz at each trial. The EMG signals were band-pass filtered between 20 and 460 Hz and differentially amplified (Input Impedance = 100GΩ, CMRR > 96 dB). Two minutes rest between support conditions and 20 s between loading levels was given to avoid fatigue during the tests.

Data analysis

To avoid transients, the middle 6 s window of the 10 s EMG signal measurement was selected for further analysis. Root mean square (RMS) values of the EMG amplitude were calculated to quantify muscle activation levels. We normalized RMS values by the use of grand maximum amount of EMG activity of each muscle within all tasks for each subject in this study. More specifically, we used maximum activation levels obtained during all trials under investigation for normalization. This method of normalizing EMG data produces high reliability between trials and greatly reduces the possibility of obtaining EMG levels during the task of interest greater than the reference value (Halaki and Ginn, 2012). Before statistical analysis, normal distribution of all EMG data was confirmed by the Shapiro—Wilk test. After computing the level of muscle contraction in four loading levels, trends of muscles were obtained and compared in different conditions. A two-way repeated measures analysis of variance (ANOVA) was performed on normalized EMG's to test the main effect of load and condition, and interaction effects of load by condition. Bonferroni pairwise comparison tests were performed to examine differences between the levels of significant independent variables. The reliability measurement was performed in two testing sessions at least 14 days apart. The reliability of all the measurements was examined using intraclass correlation coefficients with one-way ANOVA. Analyses were performed using version 17.0 of Statistical Package for the Social Sciences (SPSS) with significance level of $\alpha = 0.05$ for all tests. Sphericity of the data was confirmed by means of Mauchly's test.

Results

Assessment of reliability of EMG activity

The reliability of the EMG activity (RMS values) at different load levels and conditions was good (ICC = 0.70—0.89) to excellent (ICC $\geq$ 0.90) (Domholdt, 2005).

Myoelectric activities

A summary of results from the ANOVA with F-statistics and corresponding *p*-values is shown in Table 1. Tests of Between-Subject effect in load order effect were not significant for all muscles.

Interactive effect of load by condition on muscle EMG (RMS value)

Fig. 3 shows the interaction effect of load levels by support conditions for each muscle. Fig. 3d shows a significant main effect of condition on EMG activity of non-support side erector spinae F (2, 58) = 34.56, $p < 0.05$. EMG activity was higher in the first and second support conditions than the third condition. There was a significant effect of load F (3, 87) = 33.97, $p < 0.05$. There was also a significant interaction effect between load and condition F (6, 174) = 21.11, $p < 0.05$ which shows that EMG activity of this muscle in the two first conditions increased following the implementation of graded loading. Non-support side latissimus dorsi also showed the same pattern (Fig. 3f).

Support side erector spinae and latissimus dorsi and both IO muscles similarly showed a different pattern (Fig. 3c, e, g and h). Fig. 3c shows a significant main effect of condition on EMG activity of the support side erector spinae F (2, 58) = 51.85, $p < 0.05$. EMG activity was higher in the third condition than in the first and second conditions. There was a significant effect of load F (3, 87) = 10.29, $p < 0.05$. There was also a significant interaction effect between load and condition F (6, 174) = 40.69, $p < 0.05$ which shows that EMG activity of the support side erector spinae in the third condition increased following the implementation of graded loading. Similarly, interaction effect was significant for the support side latissimus dorsi F (6, 174) = 28.12, $p < 0.05$, the support side IO F (6, 174) = 9.27, $p < 0.05$ and the non-support side IO F (6, 174) = 10.22, $p < 0.05$. This shows that EMG activity in the third condition significantly increased in response to graded loading for support side erector spinae and latissimus dorsi and both IO muscles (Fig. 3c, e, g and h).

Under the second and third conditions, EMG activity of support side biceps femoris (Fig. 3a) increased significantly with graded loading while no changes were apparent in the first condition (interaction effect: F (6, 174) = 8.75, $p < 0.05$). In the case of non-support side biceps femoris (Fig. 3 b), EMG activity of this muscle significantly increased in response to graded loading in the third condition (interaction effect: F (6, 174) = 12.73, $p < 0.05$).

Fig. 3e and f show that by changing the load direction, an antagonistic pattern of latissimus dorsi was activated. The same pattern happened for erector spinae muscles (Fig. 3c and d). In the third condition, by increasing the task

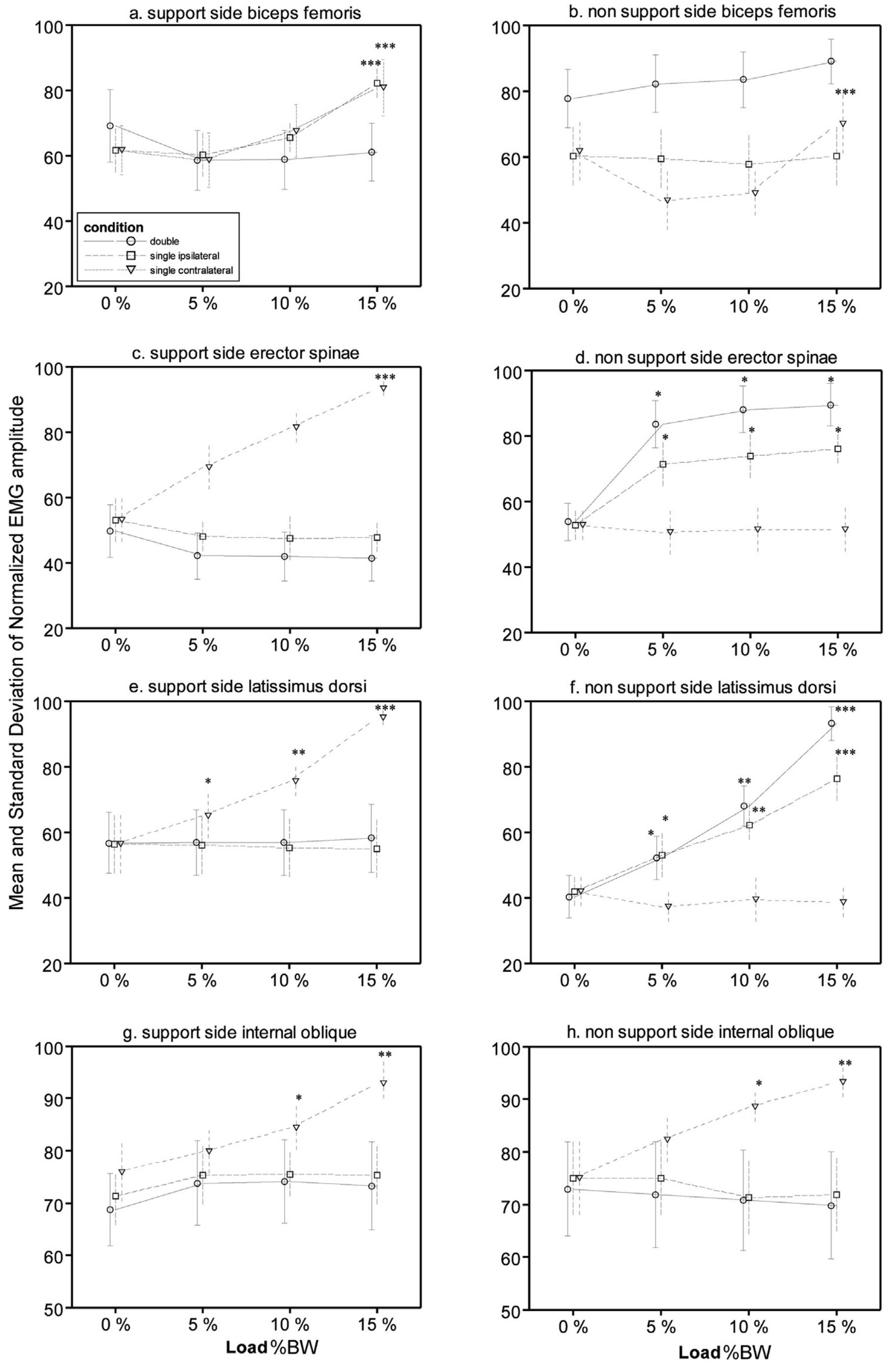

Figure 3 Interactive effect of load levels by supporting conditions on RMS value of surface electromyography signal of eight lumbopelvic muscles. Mean (SD) of the normalized RMS amplitudes of 8 lumbopelvic muscles during three different supporting conditions labelled (double: two-legged standing with the bucket in the dominant hand, Single ipsilateral: one-legged standing with the bucket in the dominant hand while using the same-side leg, and 3) single contralateral: one-legged standing using the same leg used in previous condition, but with the bucket in the non-dominant hand) and four loading levels (0%, 5%, 10% and 15% BW). Results of the Pairwise comparison ($P < 0.05$): *significantly higher than 0%BW; **significantly higher than 0%BW and 5%BW; ***significantly higher than 0%BW and 5%BW and 10%BW.

demand in one-legged standing, both IO muscles were co-activated bilaterally.

During all trials, no unilateral pattern of IO muscles was observed (Fig. 3g and h).

Discussion

Our between-days reliability test results were acceptable as it has been suggested that the reliability of between-days measures is poorer than measures taken in a single day (Yang and Winter, 1983). The higher variability in between-days reliability may be attributed to the inaccuracy of electrode repositioning and efforts variation in different days (Ng et al., 2003).

In the first condition (two-legged standing), graded loading only augmented the EMG amplitude (RMS value) of non-dominant side (contralateral to loading side) latissimus dorsi and erector spinae muscles. In the second condition (one-legged standing with ipsilateral loading), graded loading augmented the EMG amplitude of non-support side latissimus dorsi and erector spinae in addition to support side biceps femoris muscles. Finally, in the third condition (one-legged standing with contralateral loading), graded loading augmented the EMG amplitude of the support side latissimus dorsi, erector spinae, both biceps femoris and both IO muscles.

According to the conventional arch model of the pelvis, stability of the SI joints comes from *compressive* forces of transversely oriented fibers of ligaments and muscles (force closure). It appears that a one-legged (single leg support) standing position, which is analogous to the absence of one of the imposts in the arch structure, might be a defective position for the function of the SI joints when using the arch model. It is assumed that when standing on one leg, shearing load will be increased on the ipsilateral SI joint and decreased on the contralateral SI joint. In the present study, we have assessed and compared the behaviour of eight lumbopelvic muscles in the two-legged and one-legged standing positions.

In our investigation, changing position from two-legged (first condition) to one-legged (second condition) standing *did not cause any change* in the activation pattern of all trunk muscles in response to graded loading. Snijders et al. (1998) showed a decrease of contralateral IO activity when resting on one leg. They explained this decrease of IO activity by the decrease of gravity load. Although we used absolute single leg support position, we did not observe such behaviour of the IO muscles. This may suggest that changing from two-legged to one-legged standing does not alter active muscle control providing stability at the SI joints during graded loading. Additionally, in the third condition by lengthening the moment arm and consequently by the possible increase of applied load to the support side SI joint, IO activity might be expected to increase even more on the support side. However, no unilateral pattern was observed and both IOs were co-activated bilaterally. Based on these observations, it appears that a similar pattern is being used for transferring the applied load through the SI joints during all trials, regardless of the support position stance used.

In the pelvic arch model, it is hypothesized that the latissimus dorsi, due to its location in the posterior oblique sling and through its action on the posterior layer of thoracolumbar fascia, can help to compress the contralateral SI joint. At first observation, it may appear that the non-support side latissimus dorsi activates this way in the two first conditions. Interestingly enough, by changing the direction of applied load in the third condition, the activation patterns of the latissimus dorsi muscles changed, and an antagonistic pattern activated. This may suggest evidence against the hypothesis of the direct role of LD in bracing the SI joints? Bogduk et al., (1998) also showed that latissimus dorsi is designed to move the upper limb, or to raise the entire trunk in brachiation, and that its contribution to bracing the SI joints is trivial. Erector spinae muscles showed the same trend as LDs. This observation suggests that LD, together with ES, may help to control lateral bending of the trunk imposed by graded loading and to stabilize the trunk laterally (Fig. 4).

Overall, we did not observe any unilateral pattern of IO muscles (indicative of a shearing load increase or decrease at the SI joints) which may suggest that both SI joints have an equal role in load transfer — supporting positions did not make a difference.

Theoretically, an arch has to be put into *compression* to function optimally. It has to have imposts, and may also require abutments, or some other mechanism, to prevent it from collapsing, or to balance loads applied to the arch which transfer to forces pushing outward along the sides of the arch. Two-legged standing may approximate the arch analogy, but, the whole concept of an arch falls apart when a biped stands on a single leg. The 'arch' becomes a *cantilever* with completely different mechanics from a weight-bearing arch. In an arch model, increasing task demands during a higher level of activities would generate 'force closure' in the self-bracing of the SI joints, but this would require exceedingly high friction and huge musculo-ligamentous forces that are exceedingly inefficient. Additionally, all trunk muscles are part of the falling body and can only pull down (simple Newtonian physics). It is hypothesized that the piriformis muscle can come into action and increase compression forces in the SI joints because of its transverse orientation. Pel et al. (2008) performed a biomechanical analysis of SI joint stability in standing posture, using a validated static 3-D simulation model. They minimized piriformis muscle contribution to a compression force between the iliac bones and the sacrum by the simulation program, because this muscle also induces a vertical SIJ shear force. As a result, this muscle may not contribute to self-bracing mechanism.

Although further investigation is needed, the arch theory may not be biomechanically feasible as a model for pelvic stability, at least when one stands on a single leg. It appears that stability of the SI joints have not been fully recognized and, will require a more comprehensive model that can explain statics, dynamics and kinematics of this region during all functional activities. The sacrum is the connecting link to the pelvis, and should be held in place so that the spine maintains all its functions. An omnidirectional and efficient mechanical system should exist that can function in any posture and be capable of transferring

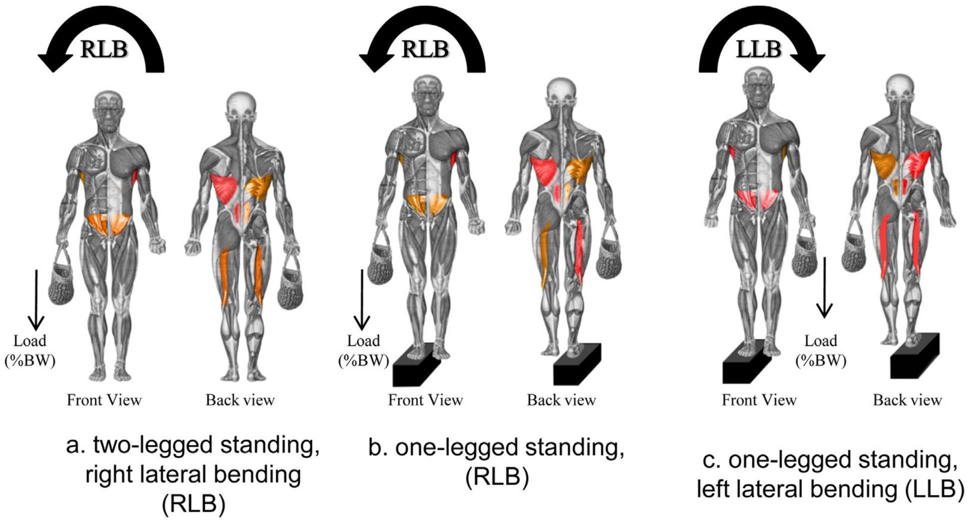

Figure 4 Panels (a—c) show global lumbopelvic muscles activation pattern during three different loading conditions. Red colour shows electrically responding muscles to graded loading in each condition. Panel a and b show external moment in the direction of right lateral bending (RLB) and panel c shows the imposed moment in the direction of left lateral bending (LLB). It indicates an antagonistic pattern of latissimus dorsi and erector spinae muscle regarding the direction of applied load. Panel c also shows the co-contraction pattern of IO muscles by increasing the moment arm of applied load. The subject depicted is assumed to be right hand dominant. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

considerable loads coming from any direction, through the pelvis and to the lower extremities.

An alternative model

Levin (2007) believes that wire-spoke bicycle wheel is a suitable model for the function of the pelvis. On a bicycle wheel, tension-loaded spokes transmit compressive loads from the frame and the ground and hub will remain suspended in its tension network. Contrary to the pillar or wagon wheel model, compression elements are secondary to the tension elements which become the primary support elements of the system (Fig. 5). Structures like this are known as "tensegrity" structures, that is a portmanteau of "tension integrity" (Fuller, 1962).

In this model, the sacrum is suspended in its network, much as the "hub" of the bicycle wheel is suspended in its spokes. According to the wire-spoke wheel model, the ligaments and muscles around the pelvis can be classified to at least three groups. First: the ligaments and muscles that have an origin on the sacrum (as hub) and an insertion on the ilia (as ring) and bridge between them. They can be considered as tension elements of a wire-spoke wheel. For instance, the sacrotuberous, sacrospinous, anterior and posterior SI joint ligaments, and the long dorsal ligament from the passive elements, and piriformis and the pelvic floor muscles from the active elements, can be assigned to this group. Secondly: those ligaments and muscles that have no direct attachment to the sacrum and connect the pelvis to the trunk and lower extremity, including the symphysis pubis and hip joint ligaments, iliolumbar ligament, abdominal muscles, quadratus lumbarum, deep hip joint flexors, extensors, adductors and abductors, can be located in this group. Thirdly: muscles that perform the movements and are located in more superficial layers. Their level of activity affects the level of tension in the two previously listed groups, whereas the proper tension state of the two previous groups is pertinent to each other, and is essential for the function of this latter group.

Some think that the situation in the body might be different because, in contrast to a bicycle wheel, the pelvic ring is not a closed ring and is open at the top and also articulated at the symphysis pubis.[6] In addition, IAP and the weight of the viscera constantly push out the ring toward the opening. This means that the pelvic rim is susceptible to opening. Accordingly, the wire-spoke wheel model's function is primarily dependent on two factors: spoke tension integrity, and rim integrity or continuity. Actually, pelvic ring integrity is the outcome of the balance between internal and external loads applied to it, which allows the sacrum to remain optimally suspended in its tension network. Based on the wire-spoke wheel model, and with regard to the aforementioned differences, those ligaments and muscles that can help to integrate the pelvic rim against *opening* have a great importance (second group). We suggest that co-contraction pattern of IO can be explained based on these assumptions.

Internal oblique muscle activation patterns

EMG activity (RMS value) of the IO muscles in the third condition was higher than its amount in the first and second conditions. This higher RMS value was *associated* with

[6] Whether the pelvic ring is open or closed is immaterial to Levin's model.

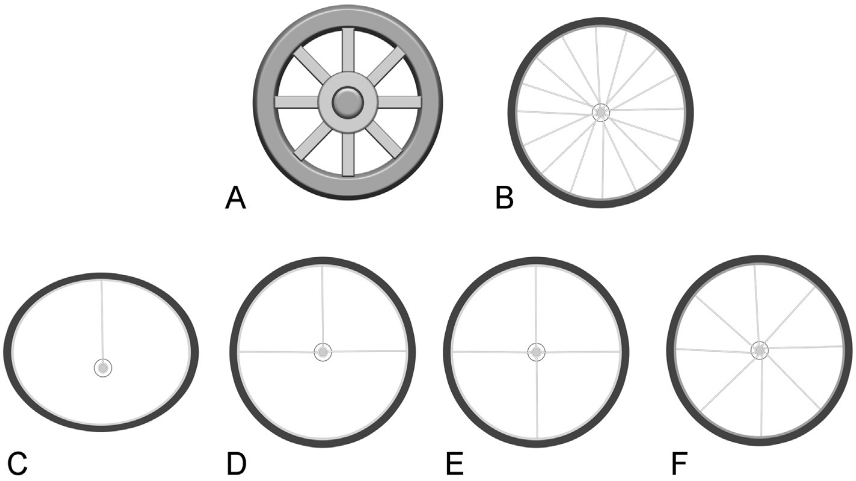

Figure 5 A. Wagon wheel. The compression spokes are thick and short. The rim is thick and heavy. The load vaults from one spoke to the next as the wheel rotates. B. Wire bicycle wheel. Long, thin tension spokes. C. The rim is thin and light and the hub is suspended by a tension spoke. D and E. It will belly out if not constrained by other tension spokes. F. Additional spokes distribute the load.
Adapted from Levin SM; the mechanics of martial arts, www.biotensegrity.com.

graded loading, meaning that increase of the load amount led to enhancement of IO muscle activity (interaction effect of load by condition). It is noticeable that this pattern occurred bilaterally indicating a co-contraction pattern between them, in this condition. Setting the correlation test between two IO muscles during 4 levels of loading confirmed the existence of a co-contraction pattern during all trials in this study. Trunk loading on the non-support side hand, increased the moment arm of the pelvis on the support leg, whereas it seems that trunk loading on the support side hand counterbalanced the consequent moment from the weight of the trunk, pelvis and lower extremity, and helped to prevent pelvic drop (Fig. 4). Accordingly, it appears that the third condition, versus two previous ones; place a higher level of task demand on the lumbopelvic stability. Regarding the wire-spoke wheel model and differences in a real situation in the body, again the third condition imposes a higher level of task demand on the SI joints which loads the pelvic rim toward shearing and opening. We therefore propose that a co-contraction pattern of the IO muscles can increase the stability of the SI joints in at least in two ways. Firstly, by opposing pelvic rim opening and help to integrate it. This is consistent with other datasets which demonstrate that spine actively achieves the stability by tuning the force in the surrounding muscles (Bergmark, 1989; Crisco and Panjabi, 1991; Cholewicki and McGill, 1995). Cholewicki and McGill (1996) observed a higher level of stability index (SI) of the lumbar spine by increasing task demands. They explained this increase in spine stability on the basis of increased muscle co-activation with increased muscular effort. Secondly, by increasing IAP, which can increase the tensional integrity and distribute it evenly through the pelvic ring. The potential role of IO in producing abdominal bracing strategies was suggested by Vera-Garcia et al. (2006). Cholewicki et al. (1999) showed that intra-abdominal pressure helps to provide stability in the lumbar spine. They observed that the stability of the spine model increased with increased IAP along with increased abdominal spring force. We suggest that the co-activation of the IO muscles might be a given strategy to increase pelvic rim integrity which is necessary for the function of the suspensory system of the sacrum. This also may suggest that respiratory dysfunctions can result in optimal weight transfer disturbance through the lumbopelvic region.

On the basis of previous studies, lumbopelvic muscles and ligaments exert forces in the direction of the SI joints bracing (force closure). We believe that this is true, not in order to increase compression in the SI joints, but for maintaining pelvic rim integrity against the opening which allows the sacrum to remain suspended in its tension network in an optimal form. The SI joint muscles and ligaments (the first group as defined in this study) are responsible for the SI joint stability as natural springs. Other structures (second and third group) help to maintain force-length relationship of SI joint ligaments and muscles and set them in a position to provide optimal pretension by tuning the pelvic ring integrity. Therapists often treat SI joint dysfunctions with overemphasis on the involved SI joints while it has been suggested that adaptations can occur symmetrically (Hungerford et al., 2003). If our proposed model holds true, one spoke cannot be adjusted without having to adjust the others. In addition, if restoring pelvic rim integrity is an essential part in the treatment, then efforts to unlock restrictions, muscular correction of positional faults and lumbopelvic, or even respiratory exercises following SI joint dysfunctions, must be taken into consideration in rehabilitation.

In the case of chronic Lumbopelvic dysfunction, maladaptations can occur which may result in sophisticated muscle responses that may result in unwanted muscle torques. Hungerford et al. (2003) observed delayed onset of IO activity at both the symptomatic and asymptomatic side in patients with SI dysfunction. Thus, in addition to their consequences, strategies can change symmetrically, which may have clinical implications for neural control treatment.

It should be noted that muscle activation patterns greatly depended on the load magnitude for each muscle. This means that during graded loading all trunk muscles responded to loading levels differently. In other words, threshold effect of the load was different for each muscle. For example, the IOs started to respond significantly to load when 10% of BW of the load was applied, while this amount was different for other muscles in this study (See Fig. 3). This observation suggests that load magnitude, and the level of task demand, must be taken into consideration when planning to design and implement spinal rehabilitation programs or studies.

Conclusions

Although further investigation is needed, it seems that the arch model is only a simplified way of possibly explaining the shape of the sacrum, and may not be biomechanically feasible for SI joints stability when one stands using single leg support. Moreover, it cannot explain muscle activation patterns during different loading conditions. Our findings are more consistent with a wire-spoke wheel model. This model is hierarchical and provides omnidirectional structural stability independent of gravity. This tension network works functionally right side up, upside down, sideways, and even when a person stands on one leg. Occurrence of muscle co-activation pattern with increased task demands appears to be a strategy to increase pelvic ring integrity. Finally, lumbopelvic muscle activation patterns are dependent on many factors, including load magnitude, direction and supporting conditions.

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