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The effect of foot wedging on electromyographic activity in the erector spinae and gluteus medius muscles during walking

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Abstract

The use of foot orthoses for treatment of low back pain (LBP) has received some attention in the literature, mainly from a clinical or theoretical perspective. It has been proposed that this treatment alleviates pain by altering muscle activity in the area of the low back but there is no direct evidence of such an effect. The objective of this study was to determine the effects of different types of foot wedging on the bilateral surface electromyographic activity of erector spinae (ErSp) (L3 level) and gluteus medius (GlMed) of 13 participants without LBP. Activity in ErSp had a significantly earlier onset during the gait cycle with bilateral heel lifts and bilateral lateral forefoot wedging. GlMed activity had a significantly later onset with bilateral heel lifts, and with an unilateral heel lift on the ipsilateral side ($P < 0.0125$). No significant amplitude changes were demonstrated in either muscle for any of the forms of wedging tested. These results show that foot wedging can produce measurable changes in timing of muscle activity within the low back and pelvis during the gait cycle. Further investigation is required to determine whether this effect contributes to the alleviation of LBP.

Keywords: Foot; Orthotic devices; Electromyography; Low back pain

1. Introduction

Low back pain (LBP) is a regular complaint of most adults worldwide. Lifetime incidence has been estimated at 60–80% of the adult population. Within the Victorian (Australian) compulsory worker's compensation scheme during 2000–2001, back injury made up the largest component of non-fatal injuries in the working population, with the cost to the community at the time in excess of A$492 million (US$270 million) per year for a population of approximately 4.5 million people [1]. Many etiologies have been implicated in the development of LBP including damage to bone, facet joints, intervertebral discs, ligaments, nerves and muscles. In addition to direct damage to these anatomical structures, the contribution of lower extremity biomechanics in the development of LBP is believed by a number of authors to be a key to understanding this pathology. The role of foot biomechanics is seen as potentially particularly important, due to the use of foot orthoses as a treatment modality to modify abnormal lower extremity biomechanics in relation to LBP [2–8].

Rothbart et al. [2] investigated 208 participants with chronic LBP, each being prescribed a pair of proprietary foot orthoses with a large medial forefoot build-up or wedge. The authors believed that almost all participants exhibited patterns of abnormal pronation, that considerably changed body posture during gait. Over 80% of participants reported at least a 50% improvement in their LBP when responding to a mailed questionnaire 1 year after treatment. A clinical study by Dananberg and Guiliano [3] involved 32 participants with chronic LBP who were unresponsive to previous traditional conservative medical management. With use of foot orthoses, participants experienced twice the improvement in pain, and for twice as long as compared to a group of participants involved in the Quebec Back Pain Disability Scale validation study [9]. Similarly, use of flat viscoe-

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lastic insoles in people with LBP was reported to result in large improvements in pain in over 80% of participants in another study at a 1-year follow-up [4]. However, since none of these three papers utilized a randomized control group, it is not clear to what extent the reported improvements were due to placebo effect.

Another approach to further research in the area will be to determine if there are measurable biomechanical changes that occur when using orthoses with people with LBP. Bird and Payne [5] reviewed theoretical mechanisms that attempt to link foot function to low back dysfunction, of which a number suggest how poor foot function might lead to the altered duration or magnitude of activity of a number of muscles proximal to the foot, including some in the low back [6–8].

Abnormal foot pronation is proposed to lead to increased internal rotation of the leg and an ipsilateral anterolateral pelvic tilt. This may then place increased strain on a number of pelvic muscles including iliopsoas, piriformis and the gluteals, subsequently leading to a rotation of the affected lumbar vertebral body, and altering dynamic forces within the low back during gait [6,7]. To decrease the stretch on iliopsoas, there may be a backward lean via active contraction of erector spinae (ErSp), theoretically resulting in muscular fatigue. The authors [6,7] hypothesize that reducing the amount of abnormal pronation with foot orthoses might resolve some cases of LBP.

The 'sagittal plane facilitation of motion' theory of foot function focuses on the ability of the sagittal plane rockers of the foot, particularly the first metatarsophalangeal joint (MTPJ), to function efficiently during gait [10]. If sagittal plane motion about these rockers is impeded during gait, this is defined as a sagittal plane blockage, and compensations within other segments of the body are predicted to occur. One theorized consequence of a functional failure of first MTPJ dorsiflexion relates to insufficient hip joint extension (involving biceps femoris contraction) at the midstance phase of the gait cycle [8]. Normally, the weightbearing limb is beginning to extend out from under the body at the midstance phase of the gait cycle. However, with a sagittal plane blockade at the first MTPJ, hip joint extension may be impeded, which may be clinically seen as a lack of normal knee joint full extension just prior to contralateral heel strike. One of the muscles responsible for hip joint extension, biceps femoris, attaches to the sacrotuberous ligament which in turn is attached to the sacrum. A lack of biceps femoris contraction is theorized to stop normal nutation of the sacroiliac joint (SIJ) whereby the sacral base should move anteriorly and the apex posteriorly. This is thought to be needed for self-bracing of the SIJ during the midstance and propulsion phases of the gait cycle [8]. Self-bracing or 'force closure' of the SIJ has also been demonstrated to occur at heel contact during the gait cycle [11]. Dananberg also hypothesizes that dysfunction of the iliopsoas and quadratus lumborum muscles are contributors to LBP development [8].

Abnormal foot pronation and failure of normal dorsiflexion at the first MTPJ during gait offer theoretical descriptions of why abnormal foot function may lead to the development of LBP, and why foot orthoses may be effective, as clinical outcome studies appear to indicate. However, the hypothetical mechanism(s) by which foot orthoses and inserts could alleviate LBP symptoms rely on a number of assumptions.

Vink and Karssemeijer [12] investigated low back muscle activity in a normal population during walking, and noted bilateral activity of the low back muscles around heel contact, presumably to counteract trunk flexion which is induced by the decelerating pelvis. An activation pattern that is usually symmetric within the muscles of the low back may be necessary to balance the forces the torso needs to withstand during walking, and it could be hypothesized that factors that provoke asymmetrical patterns over an extended period could be considered dysfunctional and may precipitate pain in the low back [13]. Vink and Huson [14] investigated the effect of raised shoe soles during treadmill walking, and noted a significant but small increase in duration of activity of the intrinsic lumbar back muscles. Parts of this protocol are deserving of a follow-up study with other modifications to foot posture, with a more clearly-defined normalization procedure for determining amplitude changes and with reliable determination of the timing of muscle activity during the gait cycle. The effect that side dominance could have on symmetry also needs to be considered during experimentation so that any observed asymmetries allow for this factor [15].

Foot orthoses vary considerably in design [16], but most types have a number of common features. To systematically analyse any potential effects on LBP with the use of orthoses, it would be desirable to determine the design parameters of foot orthoses that are most effective in altering low back biomechanical function in a normal population.

Accordingly, the aim of the present study was to investigate whether modifying foot posture with a series of different foot wedging techniques in a normal population has an effect on low back muscle activity, as a preliminary to subsequent investigations in people with LBP.

2. Methods

Surface electromyography was recorded during the gait cycle, with various types of foot wedging. Two muscles in the low back and pelvis were investigated bilaterally: ErSp (L3 level) and gluteus medius (GlMed).

Onset and amplitude of the EMG signal were analyzed at specific phases of the gait cycle.

2.1. Subjects

Thirteen healthy right-handed participants (7 female, 6 male) were recruited for this study from undergraduate students and staff at La Trobe University. Their mean age, height and weight were 22.3 (±3.4) years, 173.1 (±7.3) cm and 71.6 (±7.8) kg, respectively. They were habitually active and reported no history of disabling LBP and reported no history of lower extremity injury that might have affected their walking, for at least 6 weeks prior to the testing. The participants had no aberrations in range of motion of joints of the lower limbs, particularly within the subtalar, midtarsal or first MTPJs. The study was approved by the University's Faculty of Health Sciences Human Ethics Committee, and all participants gave their written informed consent to participate.

2.2. Equipment

Circular silver/silver chloride surface electrodes were used, with a diameter of 8 mm and a solid conducive gel covering of 18 mm diameter (3M adult red dot model 2259, 3M Health Care, Germany). Skin preparation of electrode sites involved shaving, cleaning with 70% alcohol and slightly abrading the area of skin with fine sandpaper. The skin resistance was kept below 5 kΩ, measured between the two active electrodes with an impedance meter (model CE01, Cardiometrics Pty. Ltd, Australia)

The surface electrodes were placed over bellies of ErSp and GlMed bilaterally and parallel to muscle fiber direction:

(1) ErSp was located by palpating the iliac crest of each participant bilaterally, marking the vertebral spinous process at the same height (L4) and then palpating one spinous process higher (L3). Two marks were then made with an indelible pen, 3 cm lateral to the spinous process of L3. An electrode was placed on this point, and another 3 cm directly above. The earth electrode was placed over the medial iliac crest [17]. This muscle was chosen because dysfunction of ErSp has been implicated in the development of LBP [18,19]. The third lumbar vertebral level was used since the EMG signal at this level is closely correlated (within 15% RMS difference) to that of a deeper muscle of the low back, quadratus lumborum [16], which has also been implicated in the development of LBP [10].

(2) GlMed was located by palpating the iliac crest and placing two active electrodes (3 cm apart) parallel to the muscle fibers over the proximal third of the distance between the iliac crest and the greater trochanter [20]. Care was taken to place the electrodes anterior to the gluteus maximus to minimize cross-talk with this muscle. The earth electrode was placed over the posterior iliac crest. GlMed dysfunction has also been implicated in the development of LBP [21].

The EMG signals were passed through a preamplifier (model MA-110, Motion Lab Systems, Baton Rouge, LA) to an amplifier backpack worn by the participant, passed into a filter (model MA-100/101, Motion Lab Systems), and then collected by an 8-channel analog to digital converter using CHART 3.6.1 software (PowerLab/8SP, ADInstruments Pty. Ltd, NSW, Australia). Amplifier gains were adjusted for each participant to optimize digital conversion without 'clipping' of the signal.

The frequency bandwidth was set at 20–800 Hz, with an input resistance of 100 GΩ. The signal was full-wave rectified and sampled at 1 kHz. Disposable 18-mm diameter footswitches (Motion Lab Systems) were used to differentiate intervals of stance and swing phases of the gait cycle. They were taped with hypoallergenic tape to the midpoint of the posterior heel and on the plantar aspect of the interphalangeal joint of the great toe of each foot.

To alter foot posture, a series of prefabricated foot wedges were strapped on to bare feet (the baseline measure). Although foot orthoses are almost always used inside shoes, the investigators decided not to use standardized footwear with this study, as footwear would modify foot posture considerably from barefoot. The wedges were fabricated from high density (400 durometer) ethyl vinyl acetate, and were made in a number of sizes to accommodate different sized feet. The wedges were firmly taped to the feet with zinc oxide strapping that allowed minimal movement between the foot and the wedge.

Three types of foot wedging were used:

(a) A 5° lateral forefoot wedge (Fig. 1) with distal border just proximal to the web spaces and shaped to the metatarsal parabola. Medial and lateral borders were placed in line with corresponding edges of the feet. The proximal border of the wedge was beveled to minimal thickness. Maximum thickness (8 mm) was under the fifth MTPJ. Lateral forefoot wedging is used clinically where the lateral aspect of the forefoot requires a wedge of material to balance a forefoot valgus (everted forefoot) deformity [16] and to assist with first ray plantarflexion, therefore increasing first MTPJ dorsiflexion [22].

(b) A 5° medial forefoot wedge with borders as for the lateral wedge. Maximum thickness (8 mm) was located under the first MTPJ. Medial forefoot wedging is used clinically to balance a forefoot varus (inverted forefoot) deformity [16].

(c) A heel lift (Fig. 1b) with maximum height of 2 cm at the posterior heel, that tapered off to a minimal thickness distally. A heel cup was added to the sides of

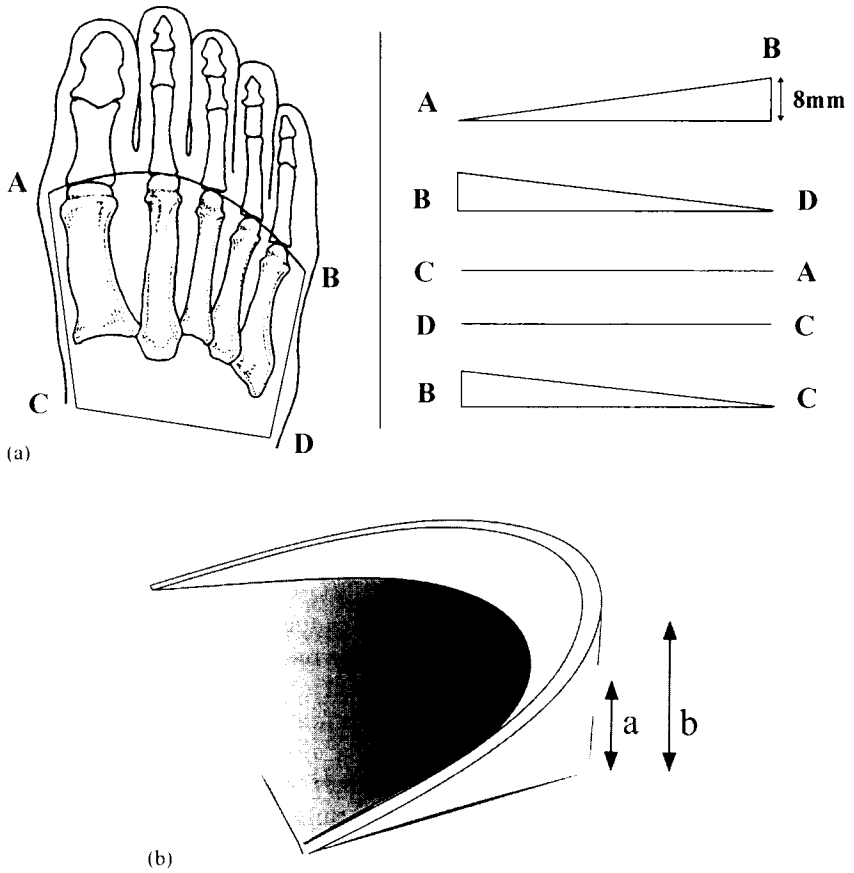

Fig. 1. (a) Diagrammatic representation of the lateral plantar forefoot wedge. The medial forefoot wedge was a mirror image of this, with maximal thickness underneath the first MTPJ. (b) A simplified diagrammatic representation of the heel lift used in this investigation. a, maximum height of heel lift (2 cm) that progressively tapered to a minimal thickness at the distal edge of this device: b, maximum height of heel cup (approximately 4 cm). The Velcro® strapping that was used to attach the heel lift to the foot is not shown.

the lift to improve participant comfort. The heel lift was attached to the foot with Velcro® strapping. Other rearfoot wedging (e.g. a medial heel wedge) was not used as it was noted in pilot testing that rearfoot wedging, other than heel lifts, tended to slide off the foot when walking without footwear. Clinically, heel lifts are used in conjunction with full foot lifts to equalize limb length discrepancies, and they have been demonstrated to significantly affect muscle activity in the low back of normal participants when walking [14] and standing [23], albeit by small amounts. Triano [23] noted that, on average, a 20 mm heel lift was necessary to affect lumbar EMG symmetry significantly when standing.

2.3. Testing procedure

Participants attended one testing session, at which they were required to wear a pair of shorts and t-shirt to enable unencumbered movement and ease of application of the electrodes and footswitches. Following initial preparation, participants performed standardized activities to allow normalization of their EMG recordings.

For GlMed, participants performed three 10 s maximum voluntary isometric contractions (MVICs) lying on their side on a padded plinth (this was completed for both sides). The investigator asked participants to perform a maximal contraction of GlMed by abducting their hip whilst he applied manual pressure to stabilize hip movement with one hand, and the ankle with the other [20]. The average EMG level for this selection was calculated by the computer, and the largest of the three values was used as the normalization level for subsequent trials, such that data recorded during experimentation was expressed as a percentage of this value.

For ErSp, one of the most common techniques used to undertake a MVIC [24] involves trunk extension when lying prone. This technique was not utilized because it was noted in preliminary testing that this was quite uncomfortable for participants, and that the signal output was significantly affected by fatigue, even with up to 3 min rest between contractions. Instead, a reference voluntary contraction was used. The modified–modified Schöber (MMS) skin distraction technique [25] uses a flexible tape-measure to measure the amount of skin distraction over the lumbar spine

between standing erect to a fully flexed position of the lumbar spine. The authors placed the trunk in a position of 20% of maximum skin distraction to obtain a consistent level of submaximal EMG. The same procedure as used for GlMed yielded a normalization level from which subsequent recordings were expressed as a percentage. This level of muscle activation was well above the baseline measure, and not affected by the flexion relaxation phenomenon [26]. The MMS technique has been shown to have Pearson Product-Moment Correlation coefficients for test-retest reliability between 0.78 and 0.89 for lumbar flexion [25].

The laboratory was set up with a 10-meter walkway, with the lightweight cabling from the EMG backpack suspended on a freely moving overhead harness. Participants were required to undertake 3 min of acclimatization to each of the foot wedge conditions and the barefoot condition by walking along the walkway. At this time, raw EMG recordings were taken to ensure that the peaks and onset of activity were reproduced and to verify that movement artifact was minimal. If the EMG signal had a wavering baseline, if there were large 'spikes' in what otherwise appeared to be normal activity or if common baseline deflections were recorded on all channels, that data was excluded from analysis.

The order in which participants undertook each of the conditions was randomized. The conditions were:

- Barefoot (baseline)
- Both feet heel lift
- Right foot heel lift
- Both feet lateral forefoot wedging
- Right foot lateral forefoot wedging
- Both feet medial forefoot wedging
- Right foot medial forefoot wedging

Participants rested for 3 min between each of the conditions. In each condition, they were required to traverse the walkway six times, only stopping long enough to turn around each time the end of the walkway was reached.

2.4. Data reduction

Time of onset and maximum amplitude of EMG activity were chosen for analysis. Altered timing and intensity of muscle activity during the gait cycle could be an important factor in the development of muscular pain, given the repetitive nature of the activity of walking. For example, significant differences in muscle onset sequencing have been noted in the quadriceps musculature of a patellofemoral pain syndrome group compared to a control group in a stepping task [27]. Maximum amplitude has been measured by other studies that have looked at the effect of foot orthoses on lower extremity muscle activity [28,29]. These parameters were also convenient because they allow straightforward assessment of whether foot wedging has any effect on muscle activity in the low back.

The CHART 3.6.1 analysis software displayed all EMG and footswitch data. To calculate onset of muscle activity, visual determination was chosen rather than a computerized automated technique, mainly due to the fact that EMG traces in this study almost always had a clear point of transition from baseline to muscle activation. In a study comparing computer-based and visual methods of determining EMG onset, Hodges and Bui [30] suggested that although a visual method appeared to produce onsets that were consistently earlier than computer-based methods, they were highly correlated ($r = 0.9999$, $P < 0.0001$). Hodges and Bui also demonstrated that in EMG traces with low levels of baseline activity there was a reduction in the difference between visual and computer-derived onsets [30].

To determine the reliability of the visual technique for this study, the principal investigator and two laypersons (who had not used the program or technique before) were tested with two trials of 50 sequences of muscle activity, in which they were required to use an on-screen pointer to indicate where they believed the onset of EMG activity to be. All three persons were blinded to the values they determined, as another investigator recorded the values from a part of the computer screen the participants could not see. The Intraclass Correlation Coefficient (ICC (2,1)) was 0.99 for the senior investigator between the two trials, and 0.98 and 0.99 for the two laypersons. The average ICC for intertester reliability was 0.99. For an average gait cycle, this level of reliability between and within examiners allowed determination of EMG onset to a mean 0.2% of the gait cycle (S.D. $\pm$0.3%). It was therefore concluded that visual determination of muscle firing onset was an appropriate and reliable analysis technique for this study.

The timing of onset of ErSp activity was measured with respect to ipsilateral heel lift, for the phase of ErSp activity that occurs during weight acceptance of the contralateral limb [31,32]. This was expressed as a percentage of the total gait cycle. For GlMed, the latency of onset of activity was measured with respect to ipsilateral heel contact, as the other period of activity seen during the gait cycle at toe off is much smaller [32]. This was also expressed as a percentage of the total gait cycle. Maximum values of the EMG amplitude were recorded, using the same gait cycle phases as for the onsets.

When averaging multiple gait cycles, a series of four gait cycle sequences can be sufficient to use with a non-pathological population [33]. However, in the present study, 10 gait cycles recorded for each condition were used for analysis, (after exclusion of any cycles with unacceptable levels of signal artifact) as the investigators

felt that this would assist in reducing variability within the averaged data. These cycles were taken from the middle one-third of the gait cycles on the walkway, when participants' walking speed was relatively consistent.

2.5. Data analysis

Data were screened for sphericity with Mauchly's test [34], and normality was assumed for the following analyses. A two-way repeated measures design was employed. The two independent variables were condition (barefoot vs. foot wedging condition) and side (left vs. right), to determine whether the individual participant's foot wedging conditions differed from the barefoot condition and whether there any side asymmetries. Therefore, a series of 24 2×2 repeated measure analyses of variance (ANOVA) were performed on the onset and amplitude data for ErSp and GlMed (Tables 1 and 2). A family-wise error adjustment [34] was used, that reduced α to 0.0125. Multiple pairwise comparisons were deemed not necessary given that each of the independent variables had only two levels for comparison. Statistical analysis was performed with the program SPSS 10.0 (SPSS Inc., USA).

Table 1
ErSp onset and amplitude ANOVA results

Condition	Onset		Amplitude	
	Significance level	Power	Significance level	Power
BF vs. BFHL				
COND[a]	0.001*	0.969	0.330	0.155
SIDE[b]	0.977	0.050	0.388	0.131
BF vs. RFHL				
COND[a]	0.569	0.084	0.054	0.501
SIDE[b]	0.345	0.148	0.425	0.199
BF vs. BFLFF				
COND[a]	0.003*	0.923	0.485	0.102
SIDE[b]	0.535	0.091	0.877	0.052
BF vs. RFLFF				
COND[a]	0.141	0.306	0.649	0.071
SIDE[b]	0.425	0.119	0.404	0.126
BF vs. BFMFF				
COND[a]	0.036	0.584	0.451	0.111
SIDE[b]	0.672	0.069	0.856	0.053
BF vs. RFMFF				
COND[a]	0.141	0.306	0.415	0.122
SIDE[b]	0.503	0.098	0.615	0.071

Abbreviations: BF, barefoot; BFHL, both feet heel lifts; RFHL, right foot heel lift; BFLFF, both feet lateral forefoot wedging; RFLFF, right foot lateral forefoot wedging; BFMFF, both feet medial forefoot wedging; RFMFF, right foot medial forefoot wedging.

* Significant at $\alpha < 0.0125$.

[a] Condition compared to barefoot.

[b] Left vs. right side for this condition.

Table 2
GlMed onset and amplitude ANOVA results

Condition	Onset		Amplitude	
	Significance level	Power	Significance level	Power
BF vs. BFHL				
COND[a]	0.000*	1.000	0.338	0.151
SIDE[b]	0.101	0.372	0.065	0.465
BF vs. RFHL				
COND[a]	0.001*	0.967	0.500	0.098
SIDE[b]	0.613	0.077	0.058	0.487
BF vs. BFLFF				
COND[a]	0.950	0.050	0.362	0.141
SIDE[b]	0.257	0.195	0.022	0.674
BF vs. RFLFF				
COND[a]	0.182	0.257	0.132	0.319
SIDE[b]	0.259	0.194	0.043	0.547
BF vs. BFMFF				
COND[a]	0.949	0.050	0.799	0.057
SIDE[b]	0.179	0.260	0.042	0.550
BF vs. RFMFF				
COND[a]	0.226	0.217	0.644	0.072
SIDE[b]	0.287	0.177	0.046	0.535

Abbreviations: BF, barefoot; BFHL, both feet heel lifts; RFHL, right foot heel lift; BFLFF, both feet lateral forefoot wedging; RFLFF, right foot lateral forefoot wedging; BFMFF, both feet medial forefoot wedging; RFMFF, right foot medial forefoot wedging.

* Significant at $\alpha < 0.0125$.

[a] Condition compared to barefoot.

[b] Left vs. right side for this condition.

3. Results

Tables 1 and 2 list results of the ANOVA tests for both onset and amplitude for each of the conditions, for ErSp and GlMed respectively.

The mean onset and amplitude times for each muscle and side are shown in Figs. 2 and 3. With ErSp onset, the heel lift condition (bilateral) produced a significant decrease in the onset time of muscle activity, as did the lateral forefoot wedging (bilateral) condition. The medial forefoot wedging (bilateral) condition approached significance and was associated with a decrease in muscle activity onset time. For the significant conditions, this corresponded to approximately 4% (or 40 ms) of the total length of the gait cycle. For the period of activity investigated, this brought the onset of activity closer to the point of heel lift within the gait cycle.

With GlMed onset, both unilateral and bilateral heel lift conditions produced a significant delay in the onset of muscle activity. This corresponded to approximately 2% (or 20 ms) of the total length of the gait cycle.

The comparison of EMG amplitude across all conditions and in both muscles revealed no significant changes.

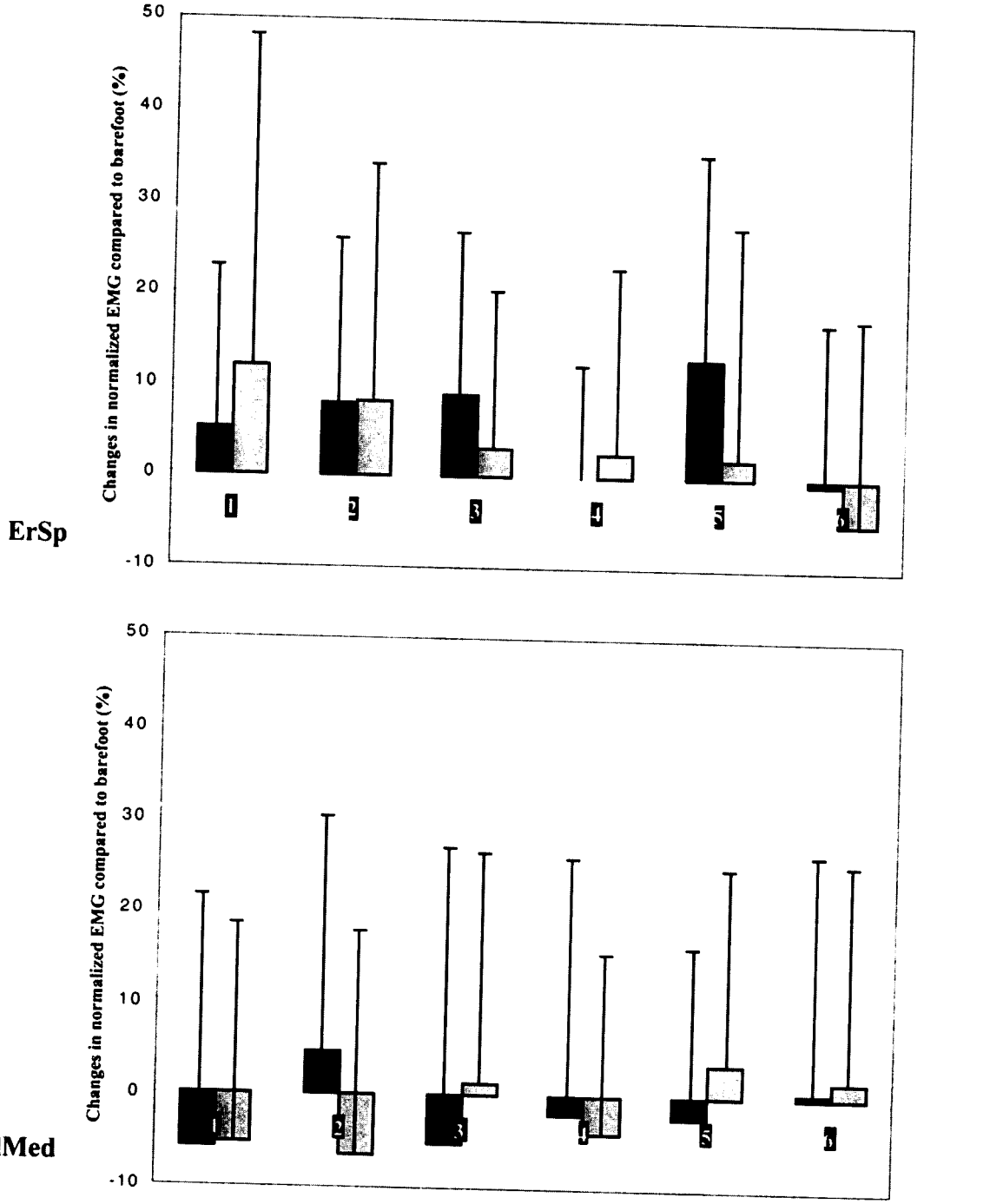

Fig. 2. Graphs of amplitude of ErSp and GlMed, indicating mean percentage changes in normalized EMG compared to barefoot, as well as showing patterns of symmetry. Black, left side; Grey, right side; 1, both feet heel lifts; 2, right foot heel lift; 3, both feet lateral forefoot wedging; 4, right foot lateral forefoot wedging; 5, both feet medial forefoot wedging; 6, right foot medial forefoot wedging. Error bars indicate 1 S.D.

4. Discussion

This present study shows that foot wedging can produce changes in muscle activity within the low back and pelvis during the gait cycle. ErSp had a significantly earlier onset of muscle activity with bilateral heel lifts and the bilateral lateral forefoot wedge conditions. In contrast, GlMed had a later onset of activity with bilateral heel lifts and was similarly affected by a unilateral heel lift on the ipsilateral side.

4.1. Effect of unilateral conditions

The only type of unilateral wedging used that had any effect was the heel lift, which altered GlMed muscle activity. This is not surprising as unilateral lifts have

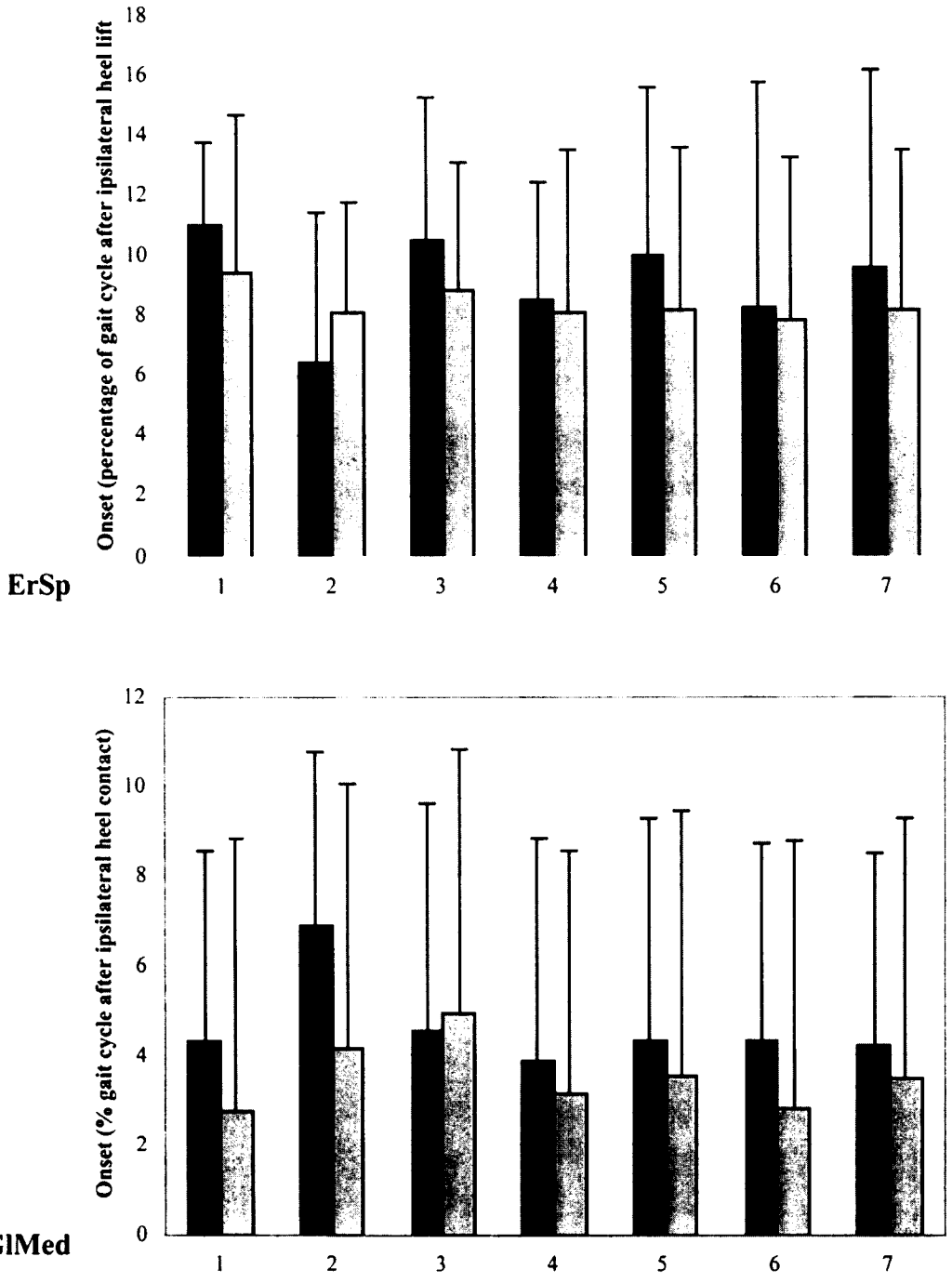

Fig. 3. Graphs of mean onset of ErSp and GlMed, showing patterns of symmetry. Black, left side; Grey, right side; 1, barefoot; 2, both feet heel lifts; 3, right foot heel lift; 4, both feet lateral forefoot wedging; 5, right foot lateral forefoot wedging; 6, both feet medial forefoot wedging; 7, right foot medial forefoot wedging. Error bars indicate 1 S.D.

been demonstrated to alter the position and tilt of the pelvis during stance [35], and GlMed principally acts as a powerful hip abductor and stabilizes pelvic tilt during the gait cycle. Triano [23] found that it took an average of 20 mm of block heel lift height (range 3–30 mm) to lead to a significant degree of 'balancing' of lumbar EMG output (visually determined) in a study involving 39 participants in static stance. However, excluding cases of limb length discrepancy, 20 mm may be uncomfortable with long term use and may be in excess of what may be required for gait rather than a static situation.

4.2. Amplitude of EMG

Although the present study took particular care in normalizing the EMG signal amplitude, no significant differences between the conditions were demonstrated for either of the muscles investigated. Similarly, no significant differences in amplitude were noted by Vink and Huson [14] who investigated 20 normal participants walking with a unilaterally raised shoe sole (not a heel lift) of 10, 20, 30 and 40 mm. As with the present study in which testing was also undertaken in a single session, Vink and Huson [14] noted large variations in EMG amplitude between participants for different conditions. As a possible way to reduce this variability, a considerably longer period of habituation to the conditions should be considered for future studies. Additionally, analyzing other aspects of the EMG signal, such as the power spectrum [36,37], may prove a more useful tool to evaluate the challenge to the muscles in question, especially in conditions that lead to muscle fatigue, which would be apparent as changes in the power spectrum [37].

The rationale for undertaking studies of EMG amplitude is that amplitude is commonly analyzed in people with LBP, and is used as a tool for clinical assessment [36,38]. The lack of significant differences in the present study may be due to use of a participant group without LBP that might show more symmetry in amplitude than people with LBP. This explanation is supported by a study which investigated the effect of inducing back pain by intramuscular injection of saline in previously unaffected participants, and recorded bilateral lumbar spine EMG amplitude from four sites during treadmill walking [39]. Significantly increased EMG amplitude was noted during the swing phase, a time at which the lumbar muscles are normally silent. The changes were located ipsilaterally to the site of pain induction. Ten other participants with pre-existing LBP also demonstrated greater ErSp activity, as well as reduced peak activity in the second period of activity during double support.

Two studies have investigated changes in leg and thigh EMG amplitude during the gait cycle with and without custom foot orthoses in normal participants [28,29]. Both studies noted considerable individual variability in the participants' response to the orthotic condition. Nawoczenski and Ludewig [29] observed only two significant changes, of which the largest was a 38% average increase in tibialis anterior amplitude in the orthotic condition compared to baseline. It would appear that, taking into account the results of these two studies and the results of the current study, EMG amplitude is not consistently altered with the introduction of an orthotic condition, at least in non-symptomatic participant groups. However, one cannot necessarily conclude that similar orthotic interventions will be ineffective in altering EMG amplitude in symptomatic participants.

4.3. Onset of EMG

Considering the whole gait cycle, the significant differences in EMG timing noted in this study are not large. The change in EMG onsets averaged approximately 4% of the gait cycle for ErSp, and 2% of the gait cycle for GlMed. However, these small timing differences may be clinically significant when it is considered that the average person repeats the gait cycle thousands of times daily. There are no studies to the authors' knowledge that have investigated timing of muscle activity during gait over an extended period, or how these changes in timing might relate to the development of pathology. However, another approach may be to consider people with paraplegia, in whom electrical stimulation can be used to elicit contraction of muscles so that such persons can perform limb movements, including walking [40,41]. Relatively precise timing of electrical stimulation is an important factor in successful use of this technique.

The changes in low back muscle activity produced by foot wedging in the present study may help to explain how orthoses could be effective in reducing LBP due to muscle dysfunction. Two mechanisms by which LBP may modulate motor performance have been discussed in the literature. Arendt-Nielsen et al. [39] suggested that modulation of muscle activity may occur via reflex pathways however, more evidence supports an assertion by Aruin and Latash of a different mechanism [42]. In their study, participants with a history of LBP undertook a series of rapid movements of the upper limb. The EMG of specific trunk muscles was significantly delayed compared to a control group. Contraction of these muscles, distant to the muscles producing movement of the limb, was described as a 'feed-forward' postural response as it occurred prior to the afferent feedback from the movement and therefore cannot be reflex in origin [42]. Further to this, Hodges and Richardson [43] compared controls and participants with LBP, where the lower limb was moved voluntarily during standing. A significant delay in EMG onset of a number of muscles of the trunk was also noted in the LBP group.

Given differences in the protocols used in these two onset studies compared to the present study, and the lack of comparison to a symptomatic group in the present study, it is not possible to make direct comparisons. However, the direction of changes in onset of activity of muscles in the present study may be clinically relevant. With ErSp in particular, the earlier onset of activity in the gait cycle achieved using foot wedging could be used as a method to counteract the later onset of trunk muscle activity in a LBP population [43]. If sequencing could be 'normalized' on a permanent basis

by foot orthoses or wedging, this could be a part of the underlying mechanism by which foot orthoses are able to reduce low back dysfunction.

4.4. Shock absorption

One factor that could have affected the results within this study is that the foot wedging could have assisted with the shock absorbing ability of the body. Ogon et al. [44] undertook a study with normal participants running barefoot and in a running shoe over a force plate overlaid with soft insole-like materials. The latency between acceleration peak (maximum external force) at the L3 level and ErSp muscle response peak (maximum internal force) was significantly longer barefoot than shod. The authors argued that the shorter latency seen with the shod condition was a mechanism to protect the lumbar spine from injury. Although not directly comparable, the results of the present study related to onset of ErSp activity with heel lifts and lateral forefoot wedging appear to show the same trend of a earlier (shorter) onset compared to barefoot. However, given that the lateral forefoot wedging could not influence shock absorption during the initial peak of ground reaction force that was investigated by Ogon et al. [44], wedging generally would appear to have properties greater than mere shock absorption.

The relative importance of the shock absorbing properties of foot orthoses compared to the design of the orthoses in modifying back muscle activity still needs to be addressed.

5. Conclusion

Foot orthoses are increasingly being used as a treatment to decrease and prevent symptoms of LBP. Their use has been met with apparent clinical success, but the underlying mechanism by which they work is unclear. The present study demonstrates that foot wedging, significantly alters onset of muscle activity in the low back and pelvis during the gait cycle. The significant interventions—heel lifts and lateral forefoot wedging—are used regularly in clinical practice as part of orthotic prescriptions. Although the results of the present study indicate that motor performance of selected low back and pelvic muscles can be altered with foot wedging, it remains to be determined whether this effect is large enough for a therapeutic benefit and whether it persists with long-term use.

References

[1] Victorian WorkCover Authority Statistical Report 2000–2001 (Australia). Melbourne: Victorian WorkCover Authority, 2001.
[2] Rothbart BA, Hansen K, Liley P, Yerratt MK. Resolving chronic low back pain: the foot connection. Am J Pain Manage 1995;5:84–90.
[3] Dananberg HJ, Guiliano M. Chronic low-back pain and its response to custommade foot orthoses. J Am Podiatr Med Assoc 1999;89:109–17.
[4] Wosk J, Voloshin AS. Low back pain: conservative treatment with artificial shock absorbers. Arch Phys Med Rehabil 1985;66:145–8.
[5] Bird AR, Payne CB. Foot function and low back pain. Foot 1999;9:175–80.
[6] Botte RR. An interpretation of the pronation syndrome and foot types of patients with low back pain. J Am Podiatr Assoc 1981;71:243–53.
[7] Minkowsky I, Minkowsky R. The spine, an integral part of the lower extremity. In: Valmassy R, editor. Clinical biomechanics of the lower extremities. St. Louis: Mosby, 1996:96–112.
[8] Dananberg HJ. Lower back pain as a gait-related repetitive motion injury. In: Vleeming A, Mooney V, Snijders CJ, Dorman TA, Stoeckart R, editors. Movement, stability and low back pain: the essential role of the pelvis. Bath: Churchill Livingstone, 1997:253–67.
[9] Kopec JA, Esdaile JM, Abrahamowicz M, Abenhaim L, Wood-Dauphinee S, Lamping DL, Williams JL. The Quebec back pain disability scale. Measurement properties. Spine 1995;20:341–52.
[10] Dananberg HJ. Functional hallux limitus and its relationship to gait efficiency. J Am Podiatr Med Assoc 1986;76:648–53.
[11] Vleeming A, Snijders CJ, Stoeckart R, Mens JMA. A new light on low back pain. In: Vleeming A, Mooney V, Dorman T, Snijders CJ, editors. Second interdisciplinary world conference on low back pain. San Diego, CA, 9–11 November 1995:149–68.
[12] Vink P, Karssemeijer N. Low back muscle activity and pelvic rotation during walking. Anat Embryol 1988;178:455–60.
[13] Solomonow M, Zhou B-H, Harris M, Lu Y, Baratta RV. The ligamento-muscular stabilizing system of the spine. Spine 1998;23:2552–62.
[14] Vink P, Huson A. Lumbar back muscle activity during walking with a leg inequality. Acta Morphol Neerl-Scand 1988;25:261–71.
[15] Merletti R, De Luca CJ, Sathyan D. Electrically evoked myoelectric signals in back muscles: effect of side dominance. J Appl Physiol 1994;77:2104–14.
[16] Michaud TC. Foot orthoses and other forms of conservative foot care, 2nd ed.. Massachusetts: Thomas C. Michaud, 1997.
[17] McGill S, Juker D, Kropf P. Appropriately placed surface EMG electrodes reflect deep muscle activity (psoas, quadratus lumborum, abdominal wall) in the lumbar spine. J Biomech 1996;29:1503–7.
[18] Hultman G, Nordin M, Saraste H, Ohlsen H. Body composition, endurance, strength, cross-sectional area, and density of MM erector spinae in men with and without low back pain. J Spinal Disord 1993;6:114–23.
[19] Lisinski P. Surface EMG in chronic low back pain. Eur Spine J 2000;9:559–62.
[20] Cram JR, Kasman GS. Introduction to surface electromyography, vol. 62–64. Maryland: Aspen Publishers Inc, 1998:352–3.
[21] Smith N. Gluteus medius function and low back pain: is there a relationship? Phys Ther Rev 1999;4:283–8.
[22] Roukis TS, Scherer PR, Anderson CF. Position of the first ray and motion of the first metatarsophalangeal joint. J Am Podiatr Med Assoc 1996;86:538–45.
[23] Triano JJ. Objective electromyographic evidence for use and effects of lift therapy. J Manipulative Physiol Ther 1983;6:13–6.
[24] Plamondon A, Marceau C, Stainton S, Desjardins P. Towards a better prescription of the prone back extension exercise to strengthen the back muscles. Scand J Med Sci Sports 1999;9:226–32.

[25] Williams R, Binkley J, Bloch R, Goldsmith CH, Minuk T. Reliability of the modified-modified Schöber and double inclinometer methods for measuring lumbar flexion and extension. Phys Ther 1993;73:26–37.

[26] Gupta A. Analyses of myo-electrical silence of erectors spinae. J Biomech 2001;34:491–6.

[27] Cowan SM, Bennell KL, Hodges PW, Crossley KM, McConnell J. Delayed onset of electromyographic activity of vastus medialis obliquus relative to vastus lateralis in subjects with patellofemoral pain syndrome. Arch Phys Med Rehabil 2001;82:183–9.

[28] Tomaro J, Burdett RG. The effects of foot orthotics on the EMG activity of selected leg muscles during gait. J Orthop Sports Phys Ther 1993;18:532–6.

[29] Nawoczenski DA, Ludewig PM. Electromyographic effects of foot orthotics on selected lower extremity muscles during running. Arch Phys Med Rehabil 1999;80:540–4.

[30] Hodges PW, Bui BH. A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography. Electroencephalogr Clin Neurophysiol 1996;101:511–9.

[31] Murray MP, Mollinger LA, Gardner GM, Sepic SB. Kinematic and EMG patterns during slow, free and fast walking. J Orthop Res 1984;2:272–80.

[32] Winter DA, Yack HJ. EMG profiles during normal human walking: stride-tostride and inter-subject variability. Electroencephalogr Clin Neurophysiol 1987;67:402–11.

[33] Yang JF, Winter DA. Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. Arch Phys Med Rehabil 1984;65:517–21.

[34] Portney LG, Watkins MP. Foundations of clinical research; applications to practice, 2nd ed.. New Jersey: Prentice Hall, 2000.

[35] Zabjek KF, Leroux MA, Coillard C, Martinez X, Griffet J, Simard G, Rivard CH. Acute postural adaptations induced by a shoe lift in idiopathic scoliosis patients. Eur Spine J 2001;10:107–13.

[36] Mohseni-Bandpei MA, Watson MJ, Richardson B. Application of surface electromyography in the assessment of low back pain: a literature review. Phys Ther Rev 2000;5:93–105.

[37] Ng JK, Richardson CA. Reliability of electromyographic power spectral analysis of back muscle endurance in healthy subjects. Arch Phys Med Rehabil 1996;77:259–64.

[38] Larivière C, Gagnon D, Loisel P. The comparison of trunk muscle EMG activation between subjects with and without chronic low back pain during flexion-extension and lateral bending tasks. J Electromyogr Kinesiol 2000;10:79–91.

[39] Arendt-Nielsen L, Graven-Nielsen T, Svarrer H, Svensson P. The influence of low back pain on muscle activity and coordination during gait: a clinical and experimental study. Pain 1995;64:231–40.

[40] Yang L, Granat MH, Paul JP, Condie DN, Rowley DI. Further development of hybrid functional electrical stimulation orthoses. Artif Organs 1997;21:183–7.

[41] Chen Y-L, Li Y-C, Kuo T-S, Lai J-S. The development of a closed-loop controlled functional electrical stimulation (FES) in gait training. J Med Eng Technol 2001;25:41–8.

[42] Aruin AS, Latash ML. Directional specificity of postural muscles in feed-forward postural reactions during fast voluntary arm movements. Exp Brain Res 1995;103:323–32.

[43] Hodges PW, Richardson CA. Delayed postural contraction of transversus abdominus in low back pain associated with movement of the lower limb. J Spinal Disord 1998;11:46–56.

[44] Ogon M, Aleksiev AR, Spratt KF, Pope MH, Saltzman CL. Footwear affects the behaviour of low back muscles when jogging. Int J Sports Med 2001;22:414–9.