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ORIGINAL RESEARCH

THE EFFECT OF A PELVIC COMPRESSION BELT ON FUNCTIONAL HAMSTRING MUSCLE ACTIVITY IN SPORTSMEN WITH AND WITHOUT PREVIOUS HAMSTRING INJURY

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ABSTRACT

Background: There is evidence that applying a pelvic compression belt (PCB) can decrease hamstring and lumbar muscle electromyographic activity and increase gluteus maximus activity in healthy women during walking. Increased isokinetic eccentric hamstring strength in the terminal range (25° - 5°) of knee extension has been reported with the use of such a belt in sportsmen with and without hamstring injuries. However, it is unknown whether wearing a pelvic belt alters activity of the hamstrings in sportsmen during walking.

Purpose: To examine the effects of wearing a PCB on electromyographic activity of the hamstring and lumbopelvic muscles during walking in sportsmen with and without hamstring injuries.

Study design: Randomised crossover, cross-sectional study.

Methods: Thirty uninjured sportsmen (23.53 $\pm$ 3.68 years) and 20 sportsmen with hamstring injuries (22.00 $\pm$ 1.45 years) sustained within the previous 12 months participated in this study. Electromyographic amplitudes of the hamstrings, gluteus maximus, gluteus medius and lumbar multifidus were monitored during defined phases of walking and normalised to maximum voluntary isometric contraction. Within-group comparisons [PCB vs. no PCB] for the normalised electromyographic amplitudes were performed for each muscle group using paired t tests. Electromyographic change scores [belt – no belt] were calculated and compared between the two groups with independent t tests.

Results: No significant change was evident in hamstring activity for either group while walking with the PCB ($p > 0.050$). However, with the PCB, gluteus medius activity ($p \leq 0.028$) increased in both groups, while gluteus maximus activity increased ($p = 0.025$) and multifidus activity decreased ($p < 0.001$) in the control group. The magnitude of change induced by the PCB in gluteus medius activity was similar between groups ($p = 0.760$). No statistically significant baseline differences in no belt scores were evident between groups for the investigated muscles ($p \geq 0.050$).

Conclusion: Application of a PCB had individual-specific effects on electromyographic activity of injured and uninjured hamstrings during walking, resulting in no significant changes within or between the two groups. Future studies investigating effects of the PCB on hamstring activity in participants with acute injury and during a more demanding functional activity such as running are warranted.

Key words: Athletic injury, hamstring, orthotic, surface electromyography, walking

Level of evidence: Level 3

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INTRODUCTION

Hamstring strain injuries are one of the most common injuries in sports such as football and rugby that involve acceleration, sprinting and kicking.[1,2] A number of risk factors for occurrence and recurrence of hamstring injuries have been considered, including age, ethnicity, prior history of hamstring injury, hamstring weakness, poor flexibility, inadequate sports-related skills, altered lumbopelvic biomechanics, and altered lumbopelvic neuromotor control.[3-5] Clinical approaches and research findings have considered the hamstrings role in eccentrically decelerating the tibia during the terminal swing phase of sprinting as a critical function in the mechanism of these injuries.[6] However, despite various treatment and preventative interventions based on eccentric strengthening, the incidence and recurrence rate of hamstring injuries have not decreased substantially over the past 20 years.

In order to advance the current understanding of factors that contribute to hamstring injuries, other functions of this muscle group require consideration. It has been argued that, in general, the hamstrings contribute to knee joint stability during stance and loading response phases,[7-8] and that biceps femoris specifically aids in tensioning the sacrotuberous ligament, helping to prepare the sacroiliac joint (SIJ) for impact during initial foot contact.[9-10] Therefore, this muscle group is also considered to have a role in stabilising the lower limb.[3] In addition, the hamstring muscles extend the hip following initial foot contact during stance phase[8,10-12] with the medial hamstrings contributing to internal rotation of the hip during the loading response phase during forward progression of the contralateral pelvis.[8] The hamstring muscles thus have considerable functional complexity which may be a factor in their high injury rates.

Peak hamstring lengthening occurs at the terminal swing phase during sprinting, when the hip is flexed (≈ 55˚to 65˚) and the knee is also slightly flexed (≈ 30˚to 45˚). Furthermore, maximal hamstring loading occurs just before peak musculotendinous stretch, and perhaps contributes to an increased injury risk during the terminal stance phase while sprinting.[13-14] A small but significant reduction in the peak hip flexion angle of hamstring-injured limbs during terminal swing phase has been observed previously, reflecting either a mechanical deficit resulting from injury or a compensatory mechanism to protect from further injury.[15] In addition, with experimentally-induced hamstring pain, gait patterns demonstrate an unloading pattern of the limb, which is apparent in decreased internal hip extensor moments present during early stance, and internal knee flexor and lateral rotator moments during the terminal stance phase.[16] However, the evidence for comparative alterations in neuromotor control of the hamstrings following injury or experimentally-induced pain during gait (walking or running) remains equivocal.[15-17] An aberrant increase in the activity of injured hamstrings during functional tasks has been argued to contribute to reinjury.[3] If application of a pelvic compression belt (PCB) can reduce electromyographic (EMG) activity of the hamstrings during functional tasks, this might provide a plausible direction for future investigation of the PCB as a treatment option for those who have sustained hamstring injury.

Application of external pelvic compression by wearing a PCB just below the anterior superior iliac spines, at the level of the pubic symphysis or above the greater trochanter has been explored as a potential intervention for patients with lumbopelvic pain, and has been shown to alter EMG recruitment patterns of the lumbopelvic and hamstring muscles.[18-20] Wearing a PCB has been hypothesized to reduce EMG activity of injured hamstrings during weight-bearing activities such as walking, based upon a number of hypothetical mechanisms related to anatomical and functional links between the hamstrings and pelvis.[21] The authors have recently reported the effects of application of a PCB on isokinetic thigh muscle strength[22] and EMG activity of the hamstrings during transition from bipedal to unipedal stance[23] in a group of sportsmen with a history of hamstring injuries. Terminal range eccentric hamstring strength was significantly increased[22] while no significant change in hamstring activity[23] was noted during transition from bipedal to unipedal stance with application of the PCB compared to the control condition. The immediate improvement in eccentric hamstring strength with application of the PCB suggests that neuromotor control influences torque production of these muscles during maximal contractions. No similar changes were evident during the weight-bearing task with submaximal functional demands.[23] Moreover, a decrease in EMG activity of the biceps femoris with application of a PCB has been reported during treadmill walking in healthy nulligravidae females[19] and during standing in both sexes.[20] Whether the application of a PCB will alter EMG activity of injured or uninjured hamstrings of sportsmen during gait (walking/running) remains unknown.

The aim of the current study was to investigate whether application of a PCB alters EMG activity of the hamstrings during over ground walking in individuals with and without hamstring injuries. In addition to the hamstrings, the lumbar multifidi and gluteal muscles were also examined by EMG analysis to better understand the changes occurring with the PCB in the lumbopelvic and proximal lower limb kinetic chain during walking.

METHODS

Study design

This was a laboratory based cross-over study in which the order of PCB conditions was randomized using computer generated numbers. Ethical approval was granted by the University of Otago Human Ethics Committee (Reference no. 11/115). All participants provided written informed consent before data collection.

Study participants

Sportsmen aged between 18 and 35 years, who participated regularly in sports at least twice weekly, were recruited in an urban setting through word of mouth, emails, flyers and adverts. As the ability to recall the occurrence of injury within the past year is reported to be reliable,[24] sportsmen were included based on their self-declaration of prior hamstring injury diagnosed by a health professional. Eligibility criteria have been described in prior published research involving the same group of sportsmen.[22,23] In brief, sportsmen with unilateral or bilateral, first-time or recurrent hamstring injury were included. A hamstring injury was defined as a sudden onset of pain in the posterior thigh during a match, competition or training session within the past year, but not less than four weeks prior to testing. Sportsmen without any previously diagnosed hamstring injury were recruited for the control group. Men with a history of diagnosis and treatment for any injury or disease of the lumbopelvic spine or lower limb (other than hamstring injury for the hamstring-injured group) within the past six months, as confirmed by clinical examination,[25] were excluded from both groups.

Electromyography

EMG data were recorded at a sampling frequency of 1500 Hz using the MyoResearch XP Master-Edition software™ 1.06.54 of a 16 channel, telemetric, Noraxon Telemyo™ 2400 T G2 system (Noraxon Inc., Scottsdale, AZ, USA). EMG signals were registered from both sides (randomly ordered for testing) for the hamstring-injured participants and left or right side (randomly selected) for the uninjured participants. The EMG active leads had an input impedance of more than 100 MΩ, a base gain of 500, and a common mode rejection ratio more than 100 dB; other properties included an input of $\pm$ 3.5 mV and a baseline noise less than 1µV RMS.

Recommendations from the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) committee were followed for skin preparation and the placement of surface electrodes (Table 1).[26-27] Two silver/silver chloride surface electrodes (Ambu® Blue Sensor SP, AMBU A/S, Denmark) were placed over the lumbar multifidus, gluteus maximus, gluteus medius, biceps femoris and semitendinosus at an inter-electrode distance of 2 cm, and the ground electrode was positioned on the spinous process of L2.

Standard manual muscle testing positions[28] were used to record EMG activity during three trials of maximum voluntary isometric contraction (MVIC) of the multifidus, gluteus maximus, gluteus medius, and hamstrings (Table 2). Once the EMG signals appeared stable on the display of EMG system's monitor, EMG was recorded for three seconds per trial for three trials. Participants rested for one minute between trials.

Motion capture

Ground reaction forces were recorded with two force plates (BP2436 and OR6-5, Advanced Medical Technologies, Newton, MA, USA), sampling at 1050 Hz. Three-dimensional kinematic data were recorded at a frequency of 100 Hz (12 EagleDigital-EGL-500RT cameras, Cortex - Motion Analysis Corporation™,

Table 1. *Guidelines for placement of surface EMG electrodes*[27]

Muscle	Site of electrode placement*	Alignment of electrodes*
Biceps femoris	Halfway between the ischial tuberosity and the lateral condyle of the tibia	Along a line joining the ischial tuberosity and the lateral condyle of the tibia
Medial hamstrings	Midway between the ischial tuberosity and the medial condyle of the tibia	In the direction of a line joining the ischial tuberosity and the medial condyle of the tibia
Gluteus maximus	At the centre of a line connecting the sacral vertebrae and the greater trochanter corresponding to the greatest prominence of the gluteal region	Along a line running from the PSIS to the posterior mid-thigh region
Gluteus medius	At 50% of the distance between the iliac crest and the greater trochanter	Along a line drawn from the iliac crest to the greater trochanter
Lumbar multifidi	2 to 3 cm lateral to the midline at the level of L5 spinous process	Along a line connecting the PSIS to the interspace between L1 and L2

*Retrieved from SENIAM guidelines (www.seniam.org).

Table 2. *Guidelines for eliciting maximum voluntary isometric contraction*[28]

Muscle	Participant's position	Researcher's position	Location of manual resistance	Movement performed
Lumbar multifidi	Prone with hands held behind the occiput	Standing to stabilise the legs using body weight and one hand	Upper thoracic spine	Extension of the lumbar spine
Gluteus maximus	Prone with knee flexed to 90˚	Standing near the pelvis on the side to be tested and stabilising the pelvis near the lumbopelvic junction with one hand	Distal part of the posterior high	Extension of the hip joint
Gluteus medius	Side lying on the contralateral side with leg to be tested uppermost	Standing behind the participant and stabilising the pelvis with one hand	Above the ankle	Abduction of the hip joint
Hamstrings	Prone lying with knee flexed between 50˚ and 25˚*	Standing near the limb to be tested	Above the ankle	Flexion of the knee joint

*Maximum voluntary contraction of the hamstrings has been reported to occur between 24˚ to 48˚.[44]

Santa Rosa, CA, USA) using a set of 23 retro-reflective skin markers.[29] To enable the use of the PCB, markers were placed over each posterior superior iliac spine (PSIS) instead of a single sacral marker. The heel, toe (head of the fifth metatarsal) and PSIS (virtual mid-PSIS) markers were used to define the gait phases. Kinematic events were analysed based on a co-ordination algorithm.[30] Orthogonal coordinate systems were aligned with the x axis pointing antero-posteriorly, the y axis pointing medio-laterally and the z axis pointing supero-inferiorly.

Pelvic compression belt

A PCB (SI-brace neoprene-ADL-anatomisch, 3200202; Rafys, The Netherlands) was manually applied below the anterior superior iliac spines (Figure 1), with maximal tension without inducing any discomfort to the participant. The mean value of PCB tension that could be achieved without self-reported discomfort in healthy male participants during walking (method similar to the present study) has been found to range between 32 N and 55 N.[31]

Walking task

Participants were asked to walk at a selected cadence of 120 steps per minute,[32] controlled by a metronome, over a 7 m walkway. A custom-made stand with two lights (one green and one red) was placed in the vicinity of the walkway (Figure 2). The light signal was used to trigger the EMG recording

Figure 1. *Application of the pelvic compression belt below the anterior superior iliac spines. This figure has been reused from the authors' previous article[23] after obtaining permission from Elsevier Limited.*

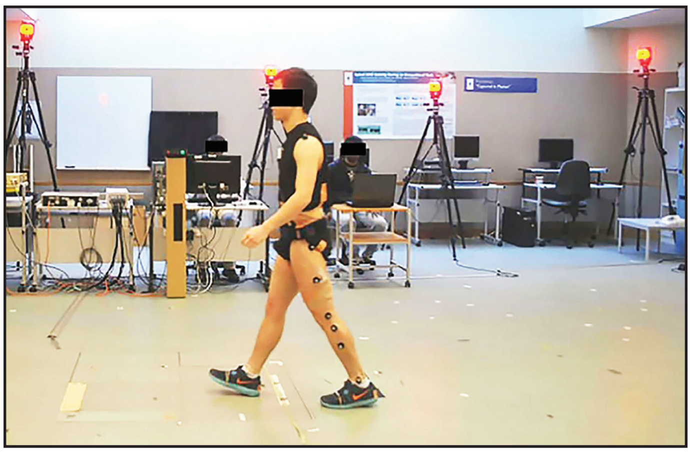

Figure 2. *Participant performing a walking trial after application of the pelvic compression belt.*

with wireless sync trigger and receiver units (234 Inline-wireless-sync-receiver and 232 Transmitter, Noraxon, Scottsdale, AZ, USA), and the participants were asked to start walking in synchrony with the metronome after the green light was switched on. Kinematic (sampling rate -100 Hz), force plate (1050 Hz) and EMG data (1500 Hz) were collected from a minimum of five successful walking trials, for each belt condition (PCB vs. no PCB) following four practice trials. Each trial lasted up to 7 s. Each successful trial consisted of walking in synchrony with the metronome beats without any obvious lag in speed and simultaneous capturing of motion with all 23 retro-reflective markers clearly visible in Cortex™ software (Version 2.0.2.917) for at least three strides. Cortex™ software was used to register the ground reaction force and light signal.

Participants also performed two other tasks with and without the PCB: 1) a bipedal to unipedal stance task[23] and 2) isokinetic strength testing of the knee.[22] Data for walking were collected following the unipedal stance task during the same session, while isokinetic testing was carried out in a separate session. This paper reports the findings of the walking task only.

Data processing

Kinematic data were processed using a Butterworth filter with a frequency cut-off of 6Hz using Cortex™ software version 2.0.2.917 (Motion Analysis Corporation, Santa Rosa, CA, USA). A pilot analysis was performed with the various methods described in Table 3 using MATLAB® software (Version 12.0.0.58851, The Mathworks, Inc.; Natick, MA, USA). The algorithm described by Zeni Jr *et al*.[30] was found to detect heel strike (HS) and toe off (TO) events more accurately than the other methods trialled for the current data. Further, this algorithm was validated with that of visually tracked HS and TO events (frame by frame analysis) with the aid of Cortex software, with or without vertical ground reaction force data for 55 gait cycles randomly selected from five participants. Pilot analyses indicated that this algorithm[30] detected HS by 10 ms ($\pm$ 7 ms) before and TO by 7 ms ($\pm$ 8 ms) after each event was identified by the visually tracked data, and was considered acceptable for the purpose of this study. Corresponding EMG data for each muscle were synchronised with the HS and TO events for each gait cycle.

EMG data were band-pass filtered within 10 to 500 Hz through a fourth order Butterworth filter. Further, the filtered EMG data were analysed with root mean square (RMS) in 50 ms epochs using MATLAB®. The mean EMG RMS value of the three MVIC trials (3 s each) for each muscle was used to normalise respective EMG amplitudes[17] and expressed as a percentage of MVIC[16] for all five muscles during the following two phases: terminal swing (20% before HS) and loading response (12% following HS). For the purpose of this study, terminal swing phase was defined as the last 20% of the gait cycle as EMG onset of the hamstrings has been reported

Table 3. *Various algorithms trialled for kinematic data in the current study to detect heel strike and toe-off events*

Algorithms used to identify HS: 1.) HS occurs at a point where the jerk (the rate of change of acceleration) is equal to zero and the acceleration is maximum for the vertical component (*z* coordinate) of the heel marker[33]. 2.) The velocity vector of horizontal (*x* coordinate) component of the heel marker changes from positive to negative at HS[30]. 3.) The timing of HS (t_{HS}) can be detected using the following formula:– $t_{HS} = (x_{Heel} - x_{Sacrum})_{maximum}$ where the sacral (virtual mid-PSIS) marker's *x* coordinate is subtracted from the heel marker's *x* coordinate at each corresponding frame and the maximum value of the resulting curve is taken as HS[30]
Algorithms used to identify TO: 1.) TO occurs at a point where the jerk is equal to zero and the acceleration is maximum for the *x* coordinate of the heel marker[33]. 2.) TO can be defined by a change in the vertical component (*z* coordinate) of the toe marker by more than 0.2cm at two consecutive time frames[34]. 3.) The velocity vector of the *x* component of the toe marker changes from negative to positive at the TO[30,35]. 4.) TO occurs when the velocity vector of the *z* coordinate of the toe marker reaches maximum[36]. 5.) Timing of TO (t_{TO}) can be detected using the following formula: $t_{TO} = (x_{Toe} - x_{Sacrum})_{minimum}$ where the sacral (virtual mid-PSIS) marker's *x* coordinate is subtracted from the toe marker's *x* coordinate at each corresponding frame and the minimum value of the resulting curve is taken as TO[30].
HS= heel strike; PSIS= posterior superior iliac spine; TO= Toe-off.

to occur during the (late) mid-swing phase.[8] Each participant walked at least five strides per trial, and the laboratory set-up allowed capture of two or three strides per trial following the first stride, depending on stride length. To counteract for acceleration and deceleration effects of walking at the start and the end of each trial respectively, only the middle stride was used for analysis. Thus, one gait cycle per trial resulting in five gait cycles per belt condition (PCB vs. no PCB) were analysed.

A customised MATLAB® program (Version 12.0.0. 58851, The Mathworks, Inc.; Natick, MA, USA) was developed to synchronize kinematic and EMG data by using the start of the light signal as the synchronizer which triggered the EMG recording. The start of the light signal was identified from the synch channel column of force plate data (analog ASCII row column file). From that point, HS and TO events were defined with tracked kinematic data (track row column file) using the alogorithm of Zeni Jr *et al.*[30] and corresponding EMG data of each muscle were synchronized for each gait cycle.

Data analysis

Terminal swing and loading response phases were included for statistical analyses of the hamstring and gluteal muscle EMG data, respectively, as maximum/most activity was observed during these periods. In addition, terminal swing was chosen as hamstring injuries are considered to occur during this phase of gait.[6] Multifidus activity was averaged during these two phases as EMG peak values were observed during HS, with initiation and cessation occurring prior to and after HS, respectively.

Data were explored for normal distribution using both the Shapiro-Wilk test and histograms, while homogeneity of variances was determined using Levene's test. Paired t tests were used to analyse within-group differences for: 1) PCB vs. no PCB trials for all the investigated muscles in each group, and 2) injured hamstring vs. uninjured hamstring EMG activity [no PCB trials] in participants with unilateral hamstring-injury. Independent *t* tests were used to 1) compare the change scores (PCB score - no PCB score) between groups and 2) to

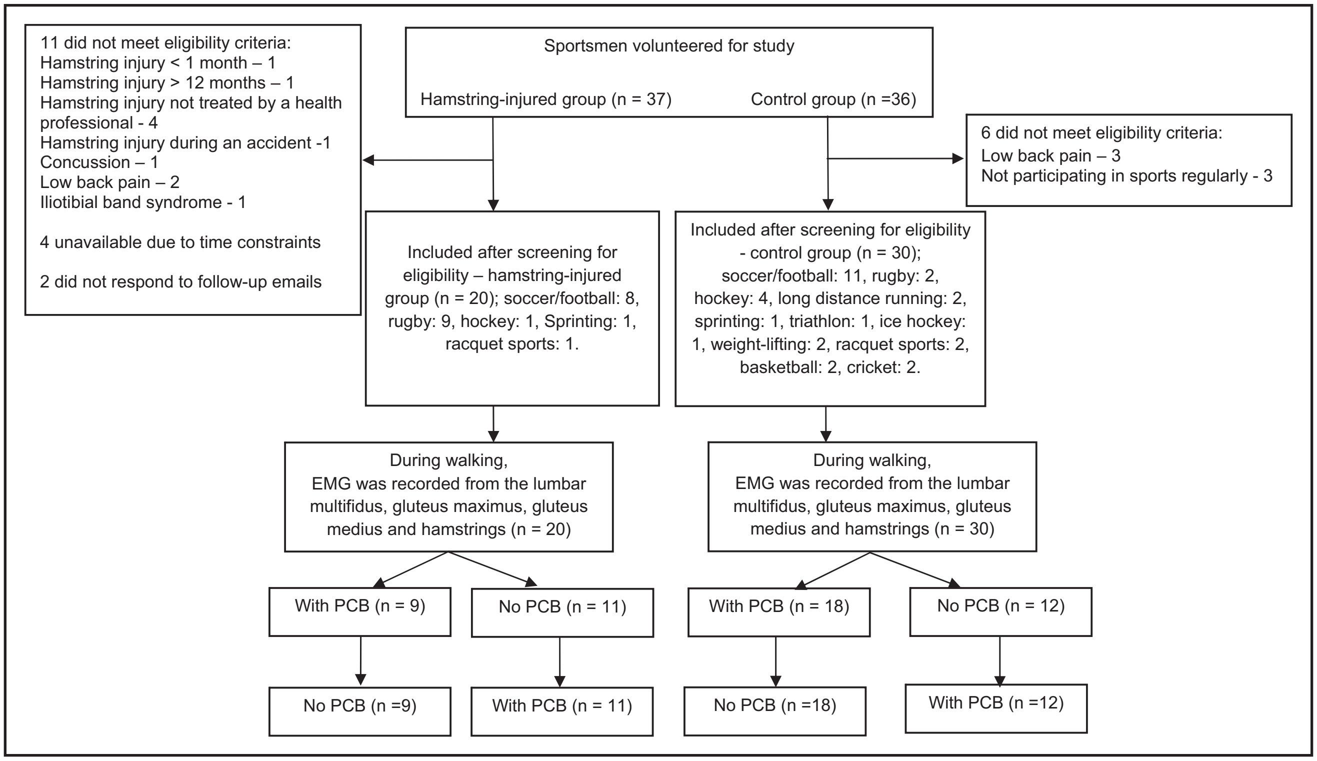

Figure 3. *Design and flow of participants through the study. In those conditions where the task was performed first with the pelvic compression belt (PCB), participants walked for 5 min after removing the belt to allow an adequate washout effect before performing the task without the belt.*

explore baseline differences in hamstring EMG activity between groups based on no PCB trials (hamstring-injured group vs. control group). Spearman's ρ correlation coefficient was used to correlate the change scores ([with PCB – no PCB]/no PCB %) of injured hamstring muscles and respective time since (recent) injury of the corresponding limb for all hamstring-injured participants. A p value < 0.05 was decided *a priori* as the level of significance.

RESULTS

A total of 37 participants for the hamstring-injured group and 36 for the healthy group volunteered between September 2011 and November 2012 for the study (Figure 3). Of these, 20 hamstring-injured volunteers (mean age, 22.00 ± 1.45 years; mean body weight, 85.52 ± 14.40 kg; mean body mass index, 25.89 ± 3.38 kg/m^2) with (n = 2) or without (n = 18) imaging investigation and 30 healthy volunteers (mean age, 23.53 ± 3.68 years; mean body weight, 70.86 ± 11.01 kg; mean body mass index, 22.92 ± 2.68 kg/m^2) fulfilled the eligibility criteria and participated in this study. Additional anthropometric characteristics of participants including height, body fat [%], and sit-and- reach values have been published previously.[23] Three participants in the hamstring-injured group had bilateral injuries and ten participants in this group had recurrent hamstring injuries. The number of injuries on the preferred side (leg self-preferred to kick a ball) and the non-preferred side were 13 and 10, respectively. The mean time since the recent injury was 4.85 (± 3.97) months and the mean time of absence from sports involvement due to injury was 3.55 (± 2.24) weeks. Four sportsmen were still undergoing some form of clinical intervention at the time of data collection. The characteristics of hamstring-injured sportsmen such as history of injury and treatment, and severity of injuries have been previously reported.[22]

A total of 22 injured limbs, including both limbs of two sportsmen with bilateral hamstring injury, and 30 uninjured limbs from healthy participants were included for analysis. One hamstring-injured limb of a participant with bilateral hamstring injury was excluded from analysis as the injury was sustained more than 12 months prior to data collection.

As data were not normally distributed, they were log transformed before conducting statistical tests.

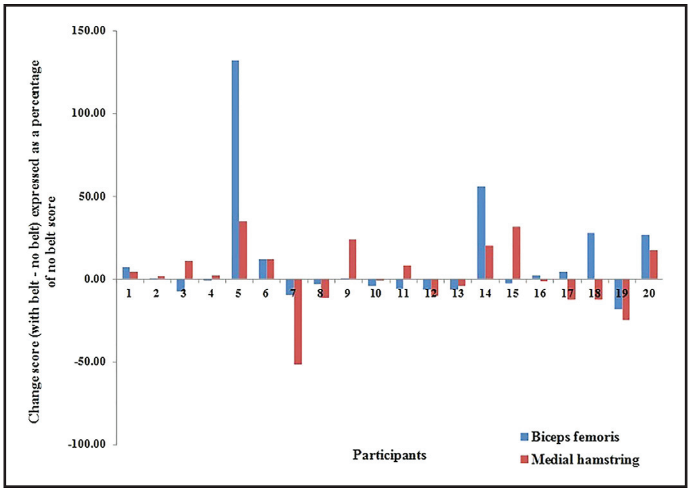

Figure 4. *Change scores (with belt - no belt) expressed as a percentage of no belt scores for normalised electromyographic values of the biceps femoris and medial hamstring muscles. Positive and negative values indicate increase and decrease in EMG values with the belt, respectively*

The log transformed data were normally distributed and so parametric statistical testing was used. The individual-specific changes for all muscle groups with application of the PCB were varied, and this is demonstrated in Figure 4, which summarises the data obtained from biceps femoris and the medial hamstrings of the injured limb.

Neither the hamstring-injured group nor the control group showed significant differences in EMG activity of the hamstrings with application of the PCB during walking (Table 4). However, there was a statistically significant difference between test conditions (PCB vs. no PCB) for MVIC normalised EMG data of the gluteus medius of the injured side of the participants in the hamstring-injured group and the multifidus, gluteus maximus and gluteus medius of the control participants during walking (Table 4). For the hamstring-injured group, only gluteus medius EMG activity significantly increased by 22% with the PCB during the loading response phase for the injured side. The mean EMG activity for the control group increased for the gluteal muscles (gluteus maximus: 16%; gluteus medius: 24%) during the loading response phase, and decreased for multifidus by 25% during the terminal swing and loading response phases with the PCB. Although both groups showed a significant increase for gluteus medius activity with the PCB, the magnitude of change induced by the PCB was not significantly different between groups ($p = 0.760$). The differences in EMG activity of multifidus and gluteus maximus obtained with and without application of the PCB were negligible for the hamstring-injured group. There was no statistically significant difference between- ($p \leq 0.569$) and within-groups ($p \leq 0.682$) for hamstring muscle EMG recorded without the PCB during the walking task. Further, the Spearman correlation between the time since injury and EMG change for the biceps femoris ($r = 0.015$, $p = 0.948$) and medial hamstrings ($r = 0.280$, $p = 0.206$) were not significant.

DISCUSSION

There was no significant difference in EMG activity of the hamstrings with application of the PCB in

Table 4. *Effects of application of the pelvic compression belt on normalized EMG RMS values of muscles of the hamstring-injured and control participants obtained during the walking task*

Muscle (MVIC normalised EMG RMS)	Gait phase	Injured limb of HIG (n = 22) Mean ± SD			Tested limb of CG (n = 30) Mean ± SD		
		No belt	With belt	P^*	No belt	With belt	P^*
Lumbar multifidus	TS + LR	17.03 ± 8.06	12.69 ± 8.54	0.146	16.05 ± 7.23	12.00 ± 5.42	**< 0.001**
Gluteus maximus	LR	14.65 ± 9.12	14.48 ± 8.45	0.332	14.11 ± 4.88	16.42 ± 6.85	**0.025**
Gluteus medius	LR	37.98 ± 14.92	46.12 ± 11.70	**0.003**	39.27 ± 18.47	48.77 ± 25.14	**0.028**
Biceps femoris	TS	23.86 ± 15.53	24.65 ± 14.29	0.078	17.57 ± 8.23	18.23 ± 8.25	0.319
Medial hamstrings	TS	33.87 ± 20.47	32.16 ± 16.89	0.678	28.65 ± 14.28	27.14 ± 13.28	0.115

Abbreviations: CG, control group; EMG RMS, electromyographic root mean square value; HIG, hamstring-injured group; LR, loading response; MVIC, maximum voluntary isometric contraction; SD, standard deviation; TS, terminal swing.

*P values based on log transformed data.

sportsmen with and without hamstring injuries. Similarly, the authors' have recently shown no significant difference for such activity during a task entailing a transition from bipedal to unipedal stance.[23] The lack of differences between the belt conditions during walking and transition from bipedal to unipedal stance[23] contrasts with other data from this group of participants whereby isokinetic eccentric hamstring strength (injured and uninjured) in the terminal range increased with the PCB.[22] Thus, while there is evidence for alteration in motor control of the hamstrings based on increased maximal eccentric strength, there appears to be no evidence for EMG changes during sub-maximal weight-bearing tasks. However, individual-specific changes with application of the PCB were noted (Figure 4). This indicates that changes in motor control of the hamstrings with the PCB could be individual- and task- specific (depending on the neuromotor demands of the task investigated) and also dependent upon the outcome variable measured.

The findings of this study contrasts with those of Hu et al[19] who reported decreased activity of biceps femoris and increased activity of gluteus maximus with the PCB during treadmill walking in healthy nulligravidae females. These results supported hypotheses proposed by other researchers for patients with low back pain.[37-38] Although both the present study and that undertaken by Hu et al[19] investigated walking, direct comparisons are limited due to various factors. Hu et al[19] calculated median EMG activity per trial (per muscle) for the whole gait cycle but their data were not normalized, nor did they randomize the order of the belt conditions (PCB vs. no PCB), meaning results could be confounded by ordering and fatigue effects. Furthermore, in contrast to the present study, Hu et al[19] included female participants and this may have also contributed towards the difference, although it is unknown how sex would contribute to this effect.[39-40]

With application of the PCB there was an increase in gluteus medius activity (22 to 25%) in both groups of participants. Park et al documented a 31% increase in gluteus medius activity in healthy participants during hip abduction in side-lying while wearing the PCB.[41] The increase noted during this non-weight bearing task appears similar to the response noted during the loading response phase (weight-bearing task) in the current study. Peak EMG activity of the gluteal muscles occurs during the loading response phase when there is increased limb loading and a subsequent need for sacroiliac joint stability.[7-8,37] Specifically, gluteus medius stabilizes the pelvis in the coronal plane, preventing the pelvis dropping on the contralateral swinging limb.[7] Application of a PCB has been hypothesized to provide proprioceptive input to facilitate gluteus medius recruitment owing to its neurophysiological (altered proprioception) and pseudofascial effects.[21] However, there was no significant difference in the magnitude of change in gluteus medius activity induced by the PCB in both groups, implying that similar neurophysiological mechanisms could have accounted for these effects in both groups.

A 25% decrease in multifidus activity with the PCB (compared to no PCB condition) during the terminal swing and loading response phases of walking in healthy participants was found. No similar changes were evident for the hamstring-injured group suggesting that this group responded differently with the PCB compared to uninjured participants. Similar to the current study, Hu et al reported an 8% decrease in erector spinae activity in healthy women (nulligravidae) while wearing the PCB during treadmill walking.[19] As both studies used similar EMG electrode placement, results appear to be similar for men and women for the low back muscles. A reduction in EMG activity of the multifidus may be due to increased force closure (forces other than the design of the articular surfaces providing sacroiliac joint stability) of the pelvic ring[19] provided by the PCB, thus necessitating less multifidi activity to augment force closure. Application of a PCB may also influence hip and knee joint angles, inducing an altered gait pattern which could affect multifidus activity. However, these putative changes require substantiation in future studies.

The heterogeneous study sample, in terms of different grades of injury and recovery, may have contributed to lack of effect of the PCB on injured hamstrings during walking. The results also indicate that the time since injury was not correlated with changes in biceps femoris and medial hamstring EMG activity with the application of the PCB. Thus, the EMG response with application of the PCB could not be predicted by the recovery phase.

This study has some limitations which require consideration. The results of this study cannot be extrapolated to the acute injury phase as all the injured sportsmen were recruited at least four weeks after injury occurrence. In addition, pain was not a major limiting factor for participants, therefore examining participants with acute hamstring injury during walking would help to determine the effects of application of the PCB on pain as well as neuromotor control. It should be noted that between group differences ($p > 0.05$) in terms of age (borderline significance), height, weight and BMI, but not for body fat measurements ($p = 0.941$) were demonstrated. However, as the mean difference in age was only 1.5 years, it is unlikely that this factor would influence EMG activity. Furthermore, as there was no significant difference between the groups for body fat measurements, it is also unlikely that the anthropometric differences would have influenced the data. Only men were investigated in the current study, thus, the findings of the study cannot be extrapolated to women. It was assumed that walking velocity was not significantly different between test conditions for both groups, though this was not objectively analyzed. As the participants walked in synchrony with a metronome for both conditions, the step length was not monitored. Thus, changes in step length affecting the EMG results would be of less significance. PCB tension was not directly measured in participants of the current study. PCB tension can change constantly during walking because of variations occurring in intra-abdominal pressure, lumbopelvic mobility and muscle activity.[31] Therefore, maintaining a constant PCB tension is not possible; however, range of values indicating optimal PCB tension that could be achieved during over ground walking was measured in a separate study on 10 healthy male participants.[31] Previous research has demonstrated that participants with pelvic girdle pain[42-43] present with altered neuromotor control of the lumbopelvic and hamstring muscles. Investigating the role of the PCB on sportspeople with more acute/subacute and severe hamstring injuries, with and without signs and symptoms of impaired pelvic stability, may help to further improve the current understanding of neuromotor control of the lumbopelvic and thigh muscles.

The neuromotor demands/control of the hamstrings during walking may be well below any threshold of detection for effect of injury on functional hamstrings muscle activity. Thus, future research may also require a similar EMG method and analysis that clearly focuses on running at various velocities in similarly injured participants and the effects of application of a PCB derived thereof.

CONCLUSION

While application of the PCB led to an increase in EMG activity of gluteus medius in participants with and without hamstring injuries, no significant effect was found for the hamstrings during the terminal swing phase of over ground walking. Therefore, such an orthotic may have limited applicability for altering hamstring activity, at least in this injured and uninjured population sample during walking. Although individual-specific responses may exist, the hamstrings may need a more demanding form of loading, or activity, in order to explore this putative phenomenon.

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