Title page

Influence of the Powers™ strap on pain and lower limb biomechanics in individuals with patellofemoral pain

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Abstract

Background: Abnormal biomechanics, especially hip internal rotation and adduction are known to be associated with patellofemoral pain (PFP). The Powers^{TM} strap was designed to decrease hip internal rotation and to thereby stabilise the patellofemoral joint.

Objectives: This study aimed to investigate whether the Powers^{TM} strap influenced pain and lower limb biomechanics during running and squatting in individuals with PFP.

Methods: 24 individuals with PFP were recruited using advertisements that were placed at fitness centres. They were asked to perform a single leg squat task (SLS) and to run on an indoor track at their own selected speed during two conditions: with and without the Powers^{TM} strap. Immediate pain was assessed with the numeric pain rating scale. Three-dimensional motion and ground reaction force data were collected with 10 Qualisys cameras and 3 AMTI force plates.

Results: Immediate pain was significantly reduced with the Powers^{TM} strap (without the Powers^{TM} strap: 4.04±1.91; with the Powers^{TM} strap: 1.93±2.13). The Powers^{TM} strap condition significantly increased hip external rotation by 4.7° during the stance phase in running and by 2.5° during the single leg squat task. Furthermore, the external knee adduction moment during the SLS and running increased significantly.

Conclusion: This study assessed the effect of the Powers™ strap on lower limbs kinematics and kinetics in individual with PFP. The results suggest that the Powers^{TM} strap has the potential to improve abnormal hip motion. Furthermore, the Powers^{TM} strap demonstrated an ability to significantly reduce pain during functional tasks in patients with PFP.

Key words: anterior knee pain, biomechanics, brace, patellofemoral pain, PFP, strap, treatment

1. Introduction

Patellofemoral pain (PFP) describes a pain around or behind the patella, which is commonly aggravated by activities that load the patellofemoral joint, such as stair stepping, squatting or running.[1] PFP is a common overuse injury that affects in particular young and physically active people and can cause limitations in performance in both sport and recreational activities.[2, 3] The pathophysiology of PFP is presumed to be multifactorial with patellofemoral malalignment and maltracking believed to play an important role in PFP. [4-7] Abnormal biomechanics, in particular dynamic knee valgus, which is a combination of hip adduction, hip internal rotation, tibial abduction and ankle eversion, are believed to be associated with patellofemoral maltracking in individuals with PFP. [8-10] Studies that have investigated the biomechanics of individuals with PFP reported an increased hip internal rotation and hip adduction angle, which was associated with higher levels of pain and reduced function in individuals with PFP [2, 3, 11-14]. Hip internal rotation leads to an inward movement of the knee joint that causes tibial abduction and foot pronation resulting in dynamic knee valgus. [28]

Abnormal lower limb biomechanics can be modified by either active interventions, such as exercise programmes and running retraining or by passive interventions, such as knee braces and patellar taping [15-19]. Passive interventions are relatively inexpensive and can be applied during sport and recreational activities [19-22]. Furthermore, a knee brace can be applied by the user without assistance from a healthcare professional and thereby can give the patient more control over the management of their PFP [23]. Several studies reported that knee braces have modified the frontal and transverse plane motion of the knee joint [24-26]. In contrast, studies investigating the influence of a passive intervention on the hip biomechanics in individuals with PFP are still lacking. The 'Powers™ strap' intends to facilitate an external rotation of the femur and thereby aims to control abnormal hip and knee motion during leisure and sport activities[27]. One study investigated the effect of the 'Powers™ strap' in healthy individuals and showed that the strap was able to effectively facilitate the external rotation of the hip during running [27]. However, only one study has investigated the influence of such a knee strap in patients with PFP during an unilateral squat and a step landing task [26]. They found that the strap significantly reduced pain and knee valgus. However, the authors measured the two-dimensional (2D) frontal-plane projection angle of the knee-valgus alignment, which did not allow the investigation of whether the strap modified the transverse plane of the hip, nor whether the strap modified lower limb kinetics [26].

Thus, the influence of the 'Powers^{TM} strap' on hip rotation and hip kinetics in individuals with PFP remains unknown. Therefore, this study aimed to investigate whether the 'Powers^{TM} strap' was able to modify hip and knee kinematics and kinetics and whether these alterations would also lead to a decrease in pain in individuals with PFP.

The Null-hypotheses were:

1. H0: The Powers^{TM} strap would not significantly decrease pain in individuals with PFP.
2. H0: There would be no significant differences in the kinematic and kinetic outcome of the hip and knee when wearing the Powers^{TM} strap in individuals with PFP.

2. Methods

The ethical approval for this study was obtained from the Salford University Ethics Research Centres Team (ERCT) (HSR 15-143) and the trial was registered at ClinicalTrials.gov (NCT02914574). Participants were recruited using advertisements that were placed at fitness centres, gyms, climbing centres and sports clubs in Manchester and Salford. Informed consent was obtained from each participant.

The eligibility criteria for individuals with PFP were: 1) aged 18-45 years; 2) antero- or retro-patellar pain with at least two of these activities: ascending or descending stairs or ramps, squatting, kneeling, prolonged sitting, hopping/ jumping, isometric quadriceps contraction or running 3) duration of current PFP symptoms >1 month.

The exclusion criteria for individuals with PFP were: (1) any history of previous lower limb surgery or patella instability and dislocation, (2) any history of traumatic, inflammatory or infectious pathology in the lower extremities or any internal derangements, including signs of effusion, (3) not able to perform running and squatting during the measurement, (4) an intake of nonsteroidal anti-inflammatory drugs

Upon the arrival a clinical assessment was carried out, which involved the Clarke’s test, a palpation test and a single leg squat task to investigate the pain region [1]. These three tests have been chosen based on the current recommendations and have shown to provide limited to good diagnostic evidence [1]. All clinical assessments were performed by the same experienced musculoskeletal physiotherapist. All participants were fitted with standard running shoes (New Balance, model M639SA UK), to control the interface of the shoe and the surface. The participants were asked to rate their pain intensity using the numeric pain rating scale (NPRS) after performing the functional tasks with and without the Powers™ strap. The instruction was “Please rate the intensity of pain on a scale of 1 to 10 that you experienced during running and the single leg squat task”. Since the application of the 3D markers and bandages might have modified the pain, the participant was also asked to rank his/her pain intensity directly after applying the bandages and markers.

2.1. 3D gait analysis

Three-dimensional (3D) movement data were collected with ten Qualisys OQUS7 cameras (Qualisys AB, Sweden) at a sampling rate of 250Hz. The 3D ground reaction forces (GRF) were collected with three force plates (BP600900, Advanced Mechanical Technology, Inc.USA), which were embedded into the floor and synchronised with the Qualisys system, at a sampling rate of 1500Hz. Forty retro-reflective markers with a diameter of 14mm were attached with double sided hypoallergic tape and bandages to the lower limbs of the participants (Figure 1). The calibrated anatomical system technique (CAST) model, which included markers on anatomical bony landmarks and segment mounted marker clusters, was used [28]. The retro-reflective markers were placed at the following anatomical landmarks: the anterior superior iliac spine, the posterior superior iliac spine, the iliac crest, the greater trochanter, the medial and lateral femoral epicondyle, the medial and lateral malleloli, the posterior calcanei, and the head of the first, second and fifth metatarsals. The four non-orthogonal tracking markers were placed on rigid clusters and were positioned over the lateral shank, and the lateral thigh of the limbs. A smaller thigh cluster was applied at the proximal thigh of the more painful limb to ensure that the Powers™ strap did not affect the cluster placement (Figure 1). A static trial was collected to specify the location of the anatomical landmark markers in relation to the clusters and to approximate the joint centre. The static trial was collected without the applied Powers™ strap but was used for both conditions with and without the Powers™ strap, because each of the marker clusters remained in the same place during both conditions.

Figure 1. The application of the markers and the Powers^{TM} strap

The participant performed all tasks firstly without and then with the applied Powers^{TM} strap which was applied on the painful knee. If both knees were affected by PFP then the Powers^{TM} strap was applied only on the more painful limb. No participant reported any adverse event due to the strap application, such as any form of discomfort or skin irritation.

2.1.1. Running task

The participant was asked to run on a 15m walkway at a self-selected speed and to walk back slowly to ensure a sufficient recovery time and to limit fatigue. Running speed was measured and reported by using Brower timing lights (Draper, UT), which were set at hip height for all participants. Each participant was asked to perform at least five running trials at a self-selected speed with five successful trials being used in the data analysis. Unsuccessful trials were the ones whereby less than three markers per segment (foot, shank, thigh, pelvis) were visible, or the foot of the focusing limb involved a partial/double foot contact with the force platforms.

2.1.2. Single leg squat task

For the performance of a single leg squat task, the participant was asked to maintain a single-leg stance on the painful leg and to fold his/her arms across his/her chest. The participants were asked to flex their knee of the non-supporting leg (approximately 90°) with no additional hip flexion (SLS-Middle). The individual was then asked to squat down as far as possible in a slow, controlled manner, while maintaining his/her balance, at a rate of approximately 1 squat per 2 seconds. The single leg squat was performed until five successful trials were recorded, whereby a trial was unsuccessful when the participants lost balance during the trial.

The participants were asked to rate his/her pain intensity using the NRPS after performing the tasks with and without the Powers™ strap.

2.2. Data processing

The kinematic and kinetic outcomes were calculated by utilising the 6-degree of freedom model in Visual3D (Version 5, C-motion Inc, USA) [27]. Marker motion data and the analogue data from the force plate were filtered with a 4th order Butterworth filter with cut-off frequencies of 12Hz. The joint kinetic outcome was calculated using three-dimensional inverse dynamics algorithm. The joint moments were normalised to body mass and presented as external moments in the local coordinate system of the proximal segment. The kinematic and kinetic data were normalised to 100% of the single leg squat and the stance phase, whereby the stance phase was sub-grouped in early stance (0-24% of stance phase), mid stance (25-62%) and late stance phase (63%-100%)[29]. The peaks of the hip and knee flexion, adduction and internal rotation angles and moments were calculated for the single leg squat and the early, mid and late stance phase in running.

2.3. Statistical analysis

The statistical analysis was performed using IBM SPSS (v. 20, IMB, USA) and Microsoft Excel 2013 (Microsoft, USA). The normality was assessed by applying the Shapiro-Wilk test and by the investigation of the normal q-q plots. For the normally distributed paired sample data, the paired t-tests were performed at the 95% confidence interval. If the data was not normal distributed and for ordinal data (pain scale) the Wilcoxon rank test was used with a significance level set at $p<0.05$.

The peak of the hip flexion, hip adduction, hip internal rotation, knee flexion, knee adduction and knee internal rotation angles and moments were compared between the conditions: with and without the Powers^{TM} strap.

The effect size for each significant variable was calculated using the Cohen d to give an indication of the magnitude of the effect of the intervention (>0.8 large effect, 0.5 moderate effect, <0.3 small effect)[30].

2.4. Power calculation

A post hoc power calculation on individuals with PFP with G-Power (Version 3.1.9.2) (n=24, one tailed t-test) was performed for all three tasks on hip internal rotation angle, by using a two-tailed t-test for two dependent means. The effect size (ES) was calculated by using the following equation (McCrum-Gardner, 2010):

$$ES = \frac{\text{(Mean of the hip IR angle with the brace)-(Mean of the hip IR angle without the brace)}}{\text{Standard deviation}}$$

The calculated effect size for the hip rotation angle in stance phase in running was d= 0.54 (medium) and thus a power of 85% was reached. The calculated effect size for the hip rotation angle during the single leg squat task was ES= 0.31 and thus only a power of 45% was achieved.

3. Results

A total of 24 individuals with PFP (12 males and 12 females, age: 29.55 ±6.44 years, height: 1.74 ± 0.09m, mass: 70.08 ±8.78kg, BMI: 23.2± 1.94) participated in the study.

The running speed of participants with PFP was on average without the Powers^{TM} strap 3.46m/s (±0.15m/s) and with the Powers^{TM} strap 3.38m/s (±0.17m/s). The speed was not significantly different between these two conditions (p=0.07).

Pain was significantly reduced with the Powers^{TM} strap during the functional tasks (p=0.0001) (without the Powers^{TM} strap: 4.04±1.91; with the Powers^{TM} strap application: 1.93±2.13, effect Cohen d: 1.04).

3.1. Running task

The hip external rotation angle was significantly increased throughout the entire stance phase when the participants were running with the Powers^{TM} strap, with an increase of hip external rotation during the: early stance phase (ESP) of 6.4°, mid stance phase (MSP) of 3.5°, late stance phase (LSP) of 4.3° (Table 1, Figure 2). However, the effect size for the early stance phase was moderate for early and small for the mid and late stance phase. The hip rotation moment increased during the early stance phase with the applied Powers^{TM} strap by 0.07Nm/kg with a moderate effect size. The knee internal rotation angle was decreased during the stance phase with a small effect size. Furthermore, the knee adduction moment was significantly increased during the stance phase. However, the effect size was small (Table 1).

Table 1. *The lower extremity kinematic and kinetic results during the stance phase in running*

The kinematic variables (º) during the stance phase in running		Without strap[1]	With strap[1]	95% Confidence Interval[2]		Std. Error of the Mean[3]	t-test, sig (2-tailed)	Effect size
				Lower	Upper			
Early stance phase	Hip flexion angle	36.3± 5.3	35.9± 5.1	-1.1	2.0	0.8	0.535	-
	Hip adduction angle	7.0± 4.6	7.3± 5.1	-2.3	1.6	1.0	0.716	-
	Hip external rotation angle	-3.2± 8.3	3.2± 8.0	4.3	8.3	1.0	0.0001†	0.79
	Knee flexion angle	31.8± 4.2	31.7± 4.1	-1.0	1.1	0.5	0.847	-
	Knee adduction angle	2.3± 4.1	1.2± 4.9	0.0	2.2	0.5	0.058	-
	Knee external rotation angle	3.2± 5.3	4.7± 5.7	0.1	2.9	0.7	0.037*	0.27
Mid stance phase	Hip flexion angle	34.5± 5.7	35.1± 5.1	-2.2	1.1	0.8	0.498	-
	Hip adduction angle	9.7±5.3	9.1± 6.8	-1.5	2.6	1.0	0.567	-
	Hip external rotation angle	1.0±8.8	4.5± 8.7	1.8	5.1	0.8	0.0002*	0.40
	Knee flexion angle	43.4± 6.3	42.5± 4.4	-1.5	3.4	1.2	0.422	-
	Knee adduction angle	-0.5± 5.0	-0.7± 5.2	-1.1	0.7	0.4	0.651	-
	Knee external rotation angle	1.9± 5.7	-0.8± 5.9	1.4	3.9	0.6	0.0002*	0.47
Late stance phase	Hip flexion angle	20.4± 5.5	21.1± 5.1	-2.2	0.8	0.7	0.330	-
	Hip adduction angle	7.2± 4.6	6.5± 5.2	-0.6	1.9	0.6	0.274	-
	Hip external rotation angle	0.2± 9.8	4.5± 10.2	2.7	5.9	0.8	0.0001*	0.43
	Knee flexion angle	41.5± 4.5	41.1±4.1	-0.7	1.5	0.5	0.501	-
	Knee adduction angle	1.0± 4.3	0.8± 4.3	-0.3	0.7	0.3	0.495	-
	Knee external rotation angle	-1.1± 5.8	1.7± 6.7	1.1	4.3	0.8	0.002†	0.45
The moment (Nm/kg) during the stance phase in running		Without strap[1]	With strap[1]	95% Confidence Interval[2]		Std. Error of the Mean[3]	t-test, sig (2-tailed)	Effect size
				Lower	Upper			
Early stance phase	Hip flexion moment	2.01± 0.44	2.00± 0.51	-0.10	0.12	0.05	0.852	-
	Hip adduction moment	1.12± 0.33	1.26± 0.45	-0.30	0.01	0.07	0.059	-
	Hip internal rotation moment	0.05± 0.10	0.12± 0.08	-0.09	-0.04	0.01	0.0001*	0.77
	Knee flexion moment	1.32± 0.49	1.43± 0.58	-0.27	0.05	0.08	0.177	-
	Knee adduction moment	0.44± 0.28	0.53± 0.33	-0.18	-0.01	0.04	0.037*	0.29
	Knee internal rotation moment	0.20± 0.11	0.25± 0.14	-0.11	0.02	0.03	0.18	-
Mid stance phase	Hip flexion moment	0.90± 0.64	0.92± 0.49	-0.25	0.23	0.12	0.919	-
	Hip adduction moment	1.82±0.45	1.84± 0.52	-0.16	0.11	0.06	0.719	-
	Hip internal rotation moment	-0.24±0.20	-0.29± 0.17	-0.03	0.12	0.03	0.198	-
	Knee flexion moment	2.41± 0.99	2.52± 0.99	-0.48	0.27	0.18	0.561	-
	Knee adduction moment	0.46± 0.32	0.57± 0.37	-0.20	-0.03	0.04	0.009*	0.32
	Knee internal rotation moment	0.41± 0.15	0.44± 0.17	-0.10	0.03	0.03	0.278	-
Late stance phase	Hip flexion moment	0.00± 0.26	-0.02± 0.28	-0.05	0.10	0.03	0.486	-
	Hip adduction moment	1.37± 0.44	1.40± 0.50	-0.14	0.08	0.05	0.586	-
	Hip internal rotation moment	0.01± 0.04	0.05± 0.11	-0.08	0.02	0.02	0.202	-
	Knee flexion moment	1.67± 0.66	1.78± 0.95	-0.44	0.21	0.16	0.478	-
	Knee adduction moment	0.31± 0.23	0.38± 0.26	-0.15	0.00	0.04	0.063	-
	Knee internal rotation moment	0.23± 0.11	0.25± 0.12	-0.06	0.01	0.02	0.204	-

*Significant ($P < .05$), [1]Mean ± standard deviation (SD), [2]95% Confidence Interval of the difference, [3]estimated SD of the sample mean

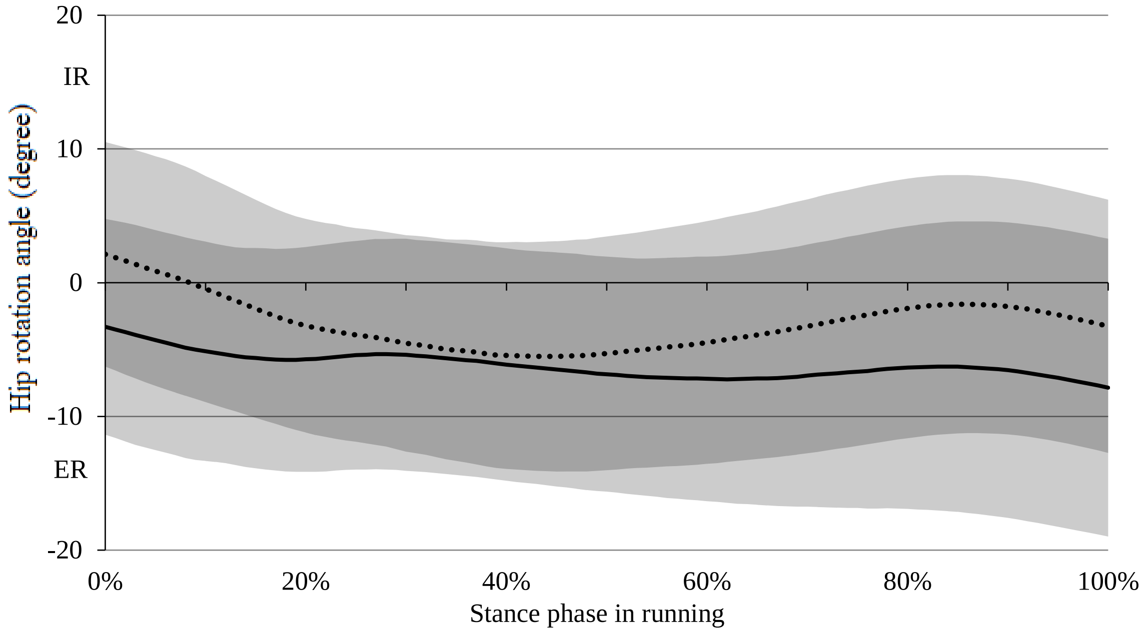

Figure 2. The hip angle in transverse plane during the stance phase of running under 2 conditions: without (dotted line) and with the Powers™ strap (solid line). The shaded areas represent ±1SD for each condition, the internal rotation angle as positive.

3.2. Single leg squat task

The hip external rotation angle significantly increased during the single leg squat task with the applied Powers™ strap (Table 2, Figure 3). Furthermore, the knee external rotation angle increased, and the hip adduction angle decreased with the applied Powers™ strap during the single leg squat task (Table 2). However, all these changes had only small effect sizes. The external knee adduction moment was significantly increased with the Powers™ strap during the single leg squat task with a moderate effect size (Table 2, Figure 4).

Table 2. The lower extremity kinematic and kinetic results during the single leg squat task

The kinematic variables (º) during the stance phase in running	Without strap[1]	With strap[1]	95% Confidence Interval[2]		Std. Error of the Mean[3]	t-test, sig (2-tailed)	Effect size
			Lower	Upper			
Hip flexion angle	73.4± 18.2	72.2± 18.3	-1.62	4.11	1.38	0.378	-
Hip adduction angle	13.6± 7.6	12.7± 7.0	0.19	1.63	0.35	0.015*	0.12
Hip external rotation angle	-0.6± 8.1	1.8± 7.6	1.48	3.33	0.45	0.0001*	0.31
Knee flexion angle	80.8± 10.7	81.0± 11.4	-2.75	2.36	1.24	0.876	-
Knee adduction angle	4.3± 4.9	4.8± 5.5	-1.28	0.24	0.37	0.172	-
Knee external rotation angle	1.4± 5.6	3.3± 5.6	0.37	3.49	0.75	0.017*	0.34
The moment (Nm/kg) during the stance phase in running	Without strap[1]	With strap[1]	95% Confidence Interval[2]		Std. Error of the Mean[3]	t-test, sig (2-tailed)	Effect size
			Lower	Upper			
Hip flexion moment	1.25± 0.58	1.25± 0.67	-0.12	0.11	0.06	0.935	-
Hip adduction moment	0.92± 0.20	0.92± 0.19	-0.05	0.04	0.02	0.821	-
Hip internal rotation moment	-0.14± 0.08	-0.13± 0.08	-0.04	0.01	0.01	0.302	-
Knee flexion moment	1.70± 0.28	1.71± 0.30	-0.07	0.05	0.03	0.689	-
Knee adduction moment	0.30± 0.10	0.36± 0.11	-0.09	-0.01	0.02	0.009*	0.57
Knee internal rotation moment	0.37± 0.09	0.39± 0.10	-0.05	0.01	0.01	0.109	-

*Significant ($P < .05$), [1]Mean ± standard deviation (SD), [2]95% Confidence Interval of the difference, [3]estimated SD of the sample mean

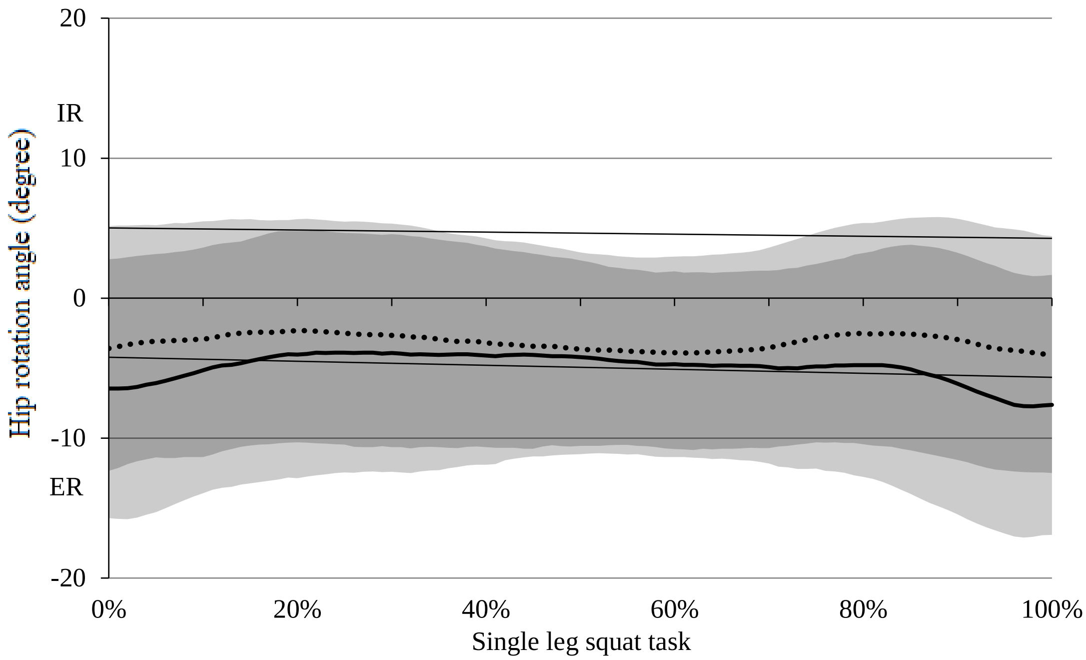

Figure 3. The hip angle in transverse plane during the single leg squat task under 2 conditions: without (dotted line) and with the Powers™ strap (solid line). The shaded areas represent ±1SD for each condition, the internal rotation angle as positive.

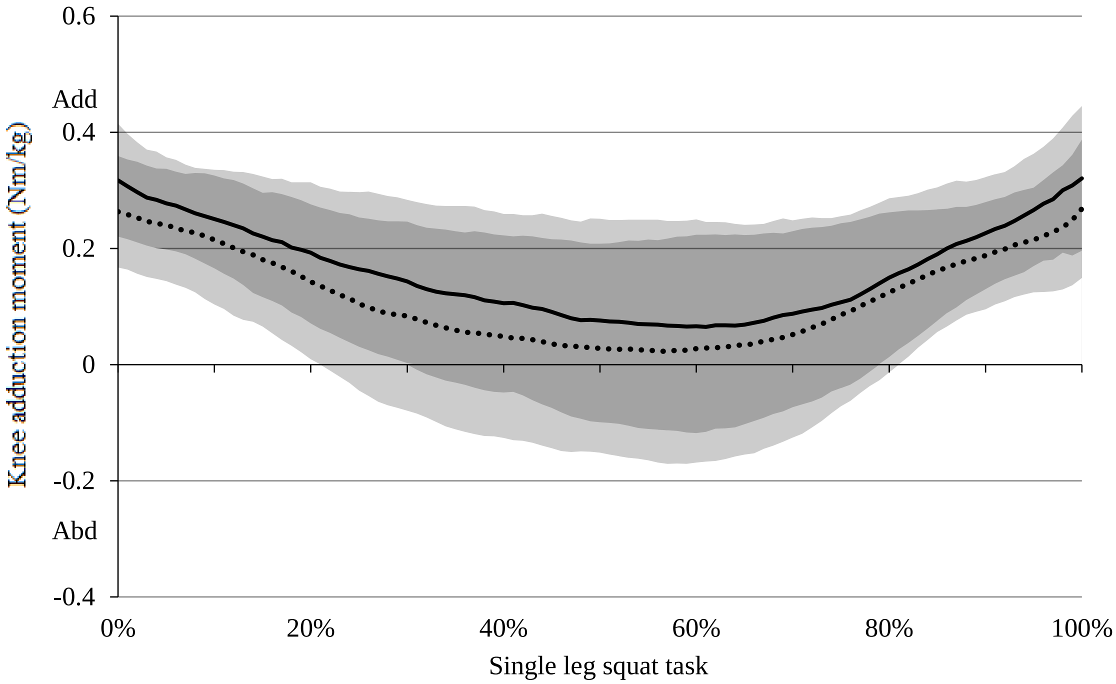

Figure 4. The knee moment in frontal plane during the stance phase of running under 2 conditions: without (dotted line) and with the Powers™ strap (solid line). The shaded areas represent ±1SD for each condition, the external adduction knee moment as positive.

4. Discussion

This study investigated hip and knee kinematics and kinetics with and without a strap of this type. The Powers™ strap significantly reduced pain with a large effect size. Pain was measured at the end of the testing battery and resulted in a drop of 2.11 in pain level after the activities with the Powers ™ strap. A clinically significant change in pain has been described as 1.74, thus the decrease of pain by 2.11 represents a clinical meaningful increase in pain [31]. Furthermore, the hip external rotation angle increased significantly during running and the single leg squat task in individuals with PFP. These findings are important because PFP can be associated with excessive hip internal rotation [13, 17, 32, 33]. Increased hip internal rotation can lead to peak patella shear stress, an increased lateral patellar tilt and displacement resulting in increased patellofemoral contact pressure [8, 34-36]. Furthermore, an increased hip internal rotation is associated with a decrease of patellofemoral contact area [36]. It is believed that a controlled hip rotation might result in decreased loading of the patellofemoral joint [14, 35]. The Powers™ strap focuses on the decrease of an increased internal rotation of the hip and appears to be a successful treatment approach.

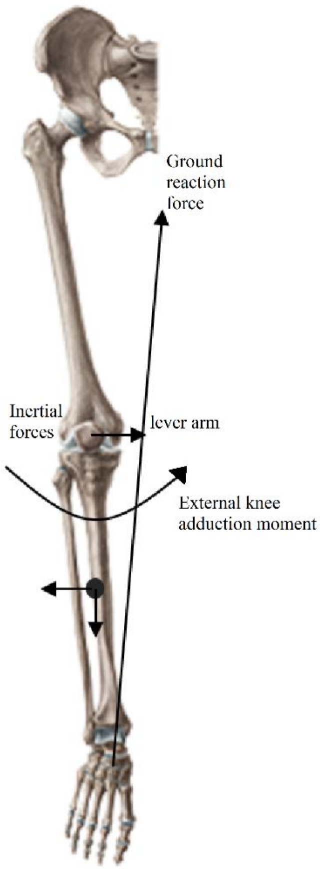

Figure 5. Diagram illustrating the external knee adduction moment during single limb stance phase [37].

The Powers™ strap also resulted in an increased knee adduction moment during the early and mid stance phase in running and the single leg squat task (Figure 5). Thus, the transverse correction of the hip resulted in a decreased dynamic knee valgus pattern. The dynamic knee valgus is characterised by an excessive hip adduction and internal rotation angle and an increased pronation of the foot [8, 11] and creates a lateral force vector on the patella that is associated to increased patellofemoral joint stress [38]. The patellofemoral joint stress reaches a peak during the early and mid stance phase [39] and thus most injuries, such as patellofemoral pain occur as a result of the high impact forces at the time of the initial contact during running [40]. The increased knee adduction moment and the decreased hip internal rotation angle during the early and mid stance phase indicate that the Powers^{TM} strap might be an effective treatment to reduce pain and effectively modifies the lower limb biomechanics in running.

To date, studies that investigated the influence of knee braces, straps and patellar taping in individuals with patellofemoral pain, concluded that bracing or taping seemed to improve acute pain, however, it did not seem to help function and stability [41-44]. This study showed that the Powers^{TM} strap reduced the acute pain significantly and had the potential to increase hip external rotation angle during running and squatting and increased the knee adduction moment. The increase of the hip external rotation angle with the Powers^{TM} strap ranged from 3.5° to 6.4°. To prove the biomechanical concept of the Powers^{TM} strap, the effect of the strap was previously investigated in 22 healthy participants and showed that the Powers^{TM} strap significantly decreased the hip internal rotation angle [27]. The reduction of the hip internal rotation angle in healthy individuals ranged between 3.2° and 4.9°, which is similar to the results in individuals with PFP. These results indicate that the Powers^{TM} strap seems to be able to influence the transverse hip biomechanics.

Although pain was significantly reduced with a large effect size, the biomechanical changes were relatively small with small to moderate effect sizes. One reason for these small changes in kinematics and kinetics might be that the individuals with PFP in this study did not show excessive hip adduction or a hip internal rotation angles and had comparable lower limb biomechanics to individuals without PFP [27]. The participants with PFP in this study were recruited from gyms and fitness centres and this recruitment strategy might have resulted in a very active and strong population of individuals with PFP. Thus, further research is required to investigate the effect of the Powers^{TM} strap in individuals with PFP that show an excessive hip internal rotation angle, though the cut off value for this has yet to be established.

Thus, this strap might be a promising treatment approach to treat patients with patellofemoral pain in acute pain and during sports activities and might enable the decrease of patellofemoral contact pressure and shear stress. However, it should be highlighted that passive interventions as a stand-alone treatment are not recommended. Instead, passive interventions, such as the Powers^{TM} strap should always be combined with exercise therapy [19, 45].

5. Methodological considerations and limitations

As with any study there are some limitations in regards to the findings of the study. It is important to note that the participants were fitted with standard training shoes to control the shoe-surface interface and to minimise the influence of footwear. However, the standard training shoes might have limited the comfort during running and thereby might have influenced the running performance. However, no individual commented that this was the case.

This study investigated the effect of the Powers™ strap within the same session and did not analyse the effect of the Powers™ strap over time. Thus, further research is required to analyse the effect of the Powers™ strap over a longer period of time to examine whether the strap might result in long-term modifications of the lower limb biomechanics and achieve a long-term pain reduction.

Individuals with PFP were not compared to healthy controls. However, the authors have previously investigated the Powers ™ strap in healthy individuals and demonstrated that the strap effectively corrected the hip internal rotation towards a neutral alignment.[16]

The authors did not investigate differences in biomechanics between females and males in this study. Thus, further research should investigate whether the Powers™ strap shows differences in biomechanics between male and female individuals with PFP.

The study investigated the application the Powers™ strap as a passive intervention. However, current guidelines for the treatment of individuals with PFP recommend the combination of passive interventions with exercises [19, 45]. Thus, further research should investigate the effect of the Powers™ strap in combination with an active exercise programme.

6. Conclusion

In conclusion, this study has demonstrated that the Powers™ strap resulted in a significant reduction of pain and was able to modify hip external rotation angle. Thus, the Powers™ strap might be a therapy to prevent excessive hip internal rotation in individuals with patellofemoral pain. However, future research should investigate the influence of the Powers™ strap over a longer period of time and should analyse the effect in individuals with PFP that show an excessive hip internal rotation angle.

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