COVER LETTER

Does the Powers^{TM} strap influence the lower limb biomechanics during running?

We are pleased to submit our manuscript entitled: "Does the Powers^{TM} strap influence the lower limb biomechanics during running?", for consideration as a full length article in gait&posture. This study is investigated whether the Powers^{TM} strap is able to modify hip kinematics and kinetics. We found that the Powers^{TM} strap significantly decreased hip and knee internal rotation in the stance phase of running."

Submission declaration and verification

The authors declare that this work has not been published previously and is not under consideration for publication elsewhere. Furthermore, all authors approve the publication of this article for gait&posture.

TITLE PAGE

Title: Does the Powers[TM] strap influence the lower limb biomechanics during running?

Authors order:
Henrike Greuel[1]
Dr. Lee Herrington[1], l.c.herrington@salford.ac.uk, office: Allerton C715
Dr. Anmin Liu[1], a.liu@salford.ac.uk, office: Brian Blatchford PO17
Prof. Richard K. Jones[1], r.k.jones@salford.ac.uk, office: Brian Blatchford PO18

1 School of Health Science, University of Salford, UK
Frederick road campus, Brian Blatchford building, PO30, Salford M66PU, Greater Manchester

Corresponding author:
Henrike Greuel[1]
School of Health Science, University of Salford, UK
Frederick road campus,
Brian Blatchford PO30,
Salford M66PU, Greater Manchester,
H.Greuel@edu.salford.ac.uk, 44 (0) 161 2952017

Does the Powers^{TM} strap influence the lower limb biomechanics during running?

Abstract

Previous research has reported a prevalence of running related injuries in 25.9% to 72% of all runners. A greater hip internal rotation and adduction during the stance phase in running has been associated with many running related injuries, such as patellofemoral pain. Researchers in the USA designed a treatment device 'the Powers^{TM} strap' to facilitate an external rotation of the femur and to thereby control abnormal hip and knee motion during leisure and sport activities. However, to date no literature exists to demonstrate whether the Powers^{TM} strap is able to reduce hip internal rotation during running.

22 healthy participants, 11 males and 11 females (age: 27.45 ±4.43 years, height: 1.73 ± 0.06m, mass: 66.77 ±9.24kg) were asked to run on a 22m track under two conditions: without and with the Powers^{TM} strap. Three-dimensional motion analysis was conducted using ten Qualisys OQUS 7 cameras (Qualisys AB, Sweden) and force data was captured with three AMTI force plates (BP600900, Advanced Mechanical Technology, Inc.USA). Paired sample t-tests were performed at the 95% confidence interval on all lower limb kinematic and kinetic data.

The Powers^{TM} strap significantly reduced hip and knee internal rotation throughout the stance phase of running. These results showed that the Powers^{TM} strap has the potential to influence hip motion during running related activities, in doing so this might be beneficial for patients with lower limb injuries. Future research should investigate the influence of the Powers^{TM} strap in subjects who suffer from running related injuries, such as patellofemoral pain.

keywords: patellofemoral pain, strap, brace, knee joint, biomechanics

Introduction:

Running is a popular sporting activity with participation rates continuing to increase [1]. Although recreational running has various beneficial health effects, it is also associated with a greater incidence of musculoskeletal injuries [1, 2]. Recent studies reported a prevalence of running related injuries in 25.9% to 72% of all runners [1, 2]. Knee injuries were prevalent in 42% of all running related injuries, followed by 16.9% for foot/ankle injuries and 12.8% for lower leg injuries. The most common overuse injury was patellofemoral pain (16.5% of all running related injuries) [3]. Studies that have investigated lower limb biomechanics in runners revealed a link between hip biomechanics and running injuries [2]. These studies showed that runners with reduced hip abductor and extensor strength exhibited greater hip internal rotation and adduction angles during the stance phase of running [4, 5]. Schmitz et al. [5] investigated the difference in running kinematics between novice and experienced runners and showed that novice runners tended towards greater hip internal rotation angles. These findings are significant because an excessive hip internal rotation can lead to lower limb injuries, such as patellofemoral pain (PFP) and non-contact anterior cruciate ligament (ACL) injuries [6]. Abnormal biomechanics, especially dynamic knee valgus, which is a combination of femoral adduction, femoral internal rotation, external knee rotation, tibial abduction, and ankle eversion is known to be associated with PFP [6, 7]. Studies that investigated the biomechanics of runners with PFP have reported an increased hip internal rotation and hip adduction compared to runners without PFP [8]. Thus, an increased hip internal rotation appears to be an associated risk factor for running associated injuries, especially patellofemoral pain.

Many different treatment options exist to modify lower limb biomechanics in runners. One commonly applied method is running retraining, which has been shown to significantly reduce the peak hip adduction [9, 10] and reduce knee adduction angles [11]. However, no significant changes on hip internal rotation have been shown with running retraining.

Insoles, knee braces and straps have also been used to modify lower limb biomechanics during running. Studies that have investigated insoles during running showed significant altered foot kinematics, but no influence on knee or hip kinematics [12]. Studies that investigated knee braces and straps for running related injuries are heterogeneous and only limited research is available [12, 13]. In addition, current research focuses on knee braces that aim to stabilise the

knee joint locally and increase proprioceptive and neuromuscular stimulation without addressing the hip movement [13, 14].

Since hip internal rotation appears to be associated with running related injuries, a brace or strap that aims to reduce the excessive hip internal rotation and thereby potentially reduce the dynamic knee valgus may be a potential treatment for running related injuries. Researchers in the USA designed a treatment device 'the Powers^{TM} strap' to facilitate an external rotation of the femur and to thereby control abnormal hip and knee motion during leisure and sport activities. 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 [15]. They found that the strap significantly reduced pain during these functional tasks and in addition significantly reduced knee valgus [15]. However, the two-dimensional (2D) frontal-plane projection angle of the knee-valgus alignment was measured and this does not allow an investigation into whether the strap modified the transverse plane movement of the hip and the knee, nor whether the strap modified lower limb kinetics. Previous studies revealed that increased knee abduction moments are associated with an increased risk for PFP and ACL injury [16-18]. Thus, individuals with lower limb injuries, such as PFP, exhibit not only altered kinematics, but also lower limb kinetics [19]. However, the influence of a knee strap that facilitate external rotation of the femur on lower limb kinetics remains unstudied.

In individuals with knee injuries, such as patellofemoral pain or after an ACL reconstruction proprioceptive deficits could be identified that were modified with knee braces and straps [20, 21]. Therefore, to ensure the investigation of the mechanistic action of the knee strap only healthy individuals were assessed.

Thus, this study aimed to investigate whether the Powers^{TM} strap is able to modify hip internal rotation angle during running in healthy individuals. Secondly, this study aimed to investigate whether the Powers^{TM} strap modified also the frontal, sagittal and transverse kinematics and kinetics of the knee and the hip during running.

The two null hypotheses were:

1. The Powers^{TM} strap would not modify the hip internal rotation angle in healthy individuals.

2. The Powers^{TM} strap would not result in changes in the frontal, sagittal and transverse kinematics and kinetics of the hip and the knee joint.

Methodology

Participants

The study was approved by the University Research and Governance Committee and the trial was registered (NCT02914574). The informed consent was obtained from each participant. Before the testing, the mass and height of each participant were measured. All participants were fitted with standard running shoes (New Balance, Abzorb soles, model M639SA UK), to control the interface of the shoe and the surface.

To be included in the study a participant had to meet all of the following criteria: (1) Active runners who have not experienced any previous significant lower limb injuries, (2) Being able to perform running, (3) in the age range: 18-45 years old. Participants were excluded if: (1) they had any history of previous lower limb surgery or patella instability and dislocation, (2) they had lower limb deformities or any history of traumatic, inflammatory or infectious pathology in the lower extremities or any internal derangements, (3) they reported previous or existing knee pain, (4) they could not perform running during the measurement.

3D gait analysis

Three-dimensional movement data were collected with ten Qualisys OQUS7 cameras (Qualisys AB, Sweden) at a sampling rate of 250Hz. The ground reaction forces were collected with three force plates (BP600900, Advanced Mechanical Technology, Inc.USA) at a sampling rate of 1500Hz, which were embedded into the floor and synchronised with the Qualisys system. Forty retro-reflective markers with a diameter of 14mm were attached, with double sided hypoallergic tape and bandages, to the lower limb of the participants (Figure 1). The calibrated anatomical system technique (CAST) model, which included anatomical landmarks (markers on anatomical bony landmarks) and anatomical frames (segment mounted marker clusters), was used [22].

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 anatomical frames were rigid clusters of 4 nonorthogonal markers 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 dominant leg to ensure that the Powers^{TM} strap could be applied below the thigh cluster and thereby did not affect the cluster placement (Figure 1). A reference trial was collected to specify the location of the anatomical landmark markers in relation to the clusters and to approximate the joint center. The ankle and knee joint centers were calculated as midpoints between the medial and lateral malleoli and femoral epicondyles, respectively. The hip joint center was calculated using the regression model of Bell [23]. A static reference trial was collected without the applied Powers^{TM} strap but was used for both conditions with and without the Powers^{TM} strap, because each of the marker clusters remained in the same place during both conditions.

Running task

Each subject was asked to run on a 22m track at his/her own selected speed during two conditions: without and with the Powers^{TM} strap. Running speed was controlled and reported by using Brower timing lights (Draper, UT) to ensure that each trial was within ±5% of the original self selected speed. The Powers^{TM} strap was applied on the preferred limb by the same researcher each time, whereby the limb dominancy was established by asking participants which limb they would prefer to kick a ball. Whilst there might be minor changes of the tightness within the individuals, this was controlled as much as possible. Furthermore, the principal researcher was experienced with the application and therefore we would hope that a standardised tightness of the strap was achieved.

Each running task was performed until five successful trials were collected. Unsuccessful trials were ones whereby less than three markers per segment were visible, running speed was out of the control range, or a partial/double contact with the force platforms was found.

Data processing

The kinematic and kinetic outcomes were calculated by utilising the 6 degrees of freedom model in Visual 3D (Version 5, C-motion Inc, USA). Motion and force plate data were filtered with a 4th order Butterworth filter with cut-off frequencies of 12Hz. The force and movement have been filtered with the same filter because studies revealed a significant effect of filtering on joint moments, especially when different cut-off frequencies had been chosen for movement and force [24]. Thus, they strongly recommended that kinetic and kinematic data should be processed with the same filter [24]. The Cardan sequence used in the kinematics calculation with Visual3D was the ordered sequence of rotations (x, y, z), with: x = flexion/extension, y = abduction/adduction, z = internal/external rotation [25]. The reliability of the applied 3D gait analysis model has been previously investigated and proved to be moderate to highly reliable during running [26].

The joint kinetic data was calculated using three dimensional inverse dynamics. The joint moments were normalised to body mass and presented as external moments referenced to the proximal segment. The kinematic and kinetic data were normalised to 100% of the stance phase, whereby the stance phase was sub-grouped in early (0-24% of stance phase), mid (25-62%) and late stance phase (63%-100%) [27]. The peaks of the hip and knee flexion, adduction and internal rotation angles and the moments were selected from the individual trials before averaging and were calculated in early, mid and late stance phase.

Statistical analysis

The statistical analysis was performed using SPSS (v. 20) and Excel 2013. The normality was assessed by applying the Shapiro-Wilk test and by the investigation of the normal q-q plots. After confirming the data met the assumption of normality, a series of paired samples t-tests were performed with the risk of Type I error at .05. The peak of the hip internal rotation angle and moments, as well as the peak of the hip flexion, hip adduction, knee flexion, knee adduction and knee internal rotation angles and moments during early, mid and late stance phase were compared between with and without the Powers[TM] strap.

6

Results

Twenty-two healthy participants, 11 males and 11 females (age: 27.45 ±4.43 years, height: 1.73 ± 0.06m, mass: 66.77 ±9.24kg) participated in the study.

The participants' running speed on average without the Powers[TM] strap was 3.4m/s (±0.3m/s) and with the Powers[TM] strap 3.3m/s (±0.2m/s). The speed was not significantly different between the two conditions (p=0.08).

The hip internal rotation angle (Figure 2) decreased by 3.2° during early stance phase (p=0.011), 3.4° during mid stance phase (p=0.001) and 4.9° during late stance phase (p=0.0001) when the participants were running with the Powers[TM] strap. Additionally, the knee internal rotation angle (Figure 3) decreased during the early stance by 1.6° (p=0.025) and mid stance phase by 2.0° (p=0.002) in running with the Powers[TM] strap, but not in late stance. The knee adduction angle decreased by 0.9° (p=0.034) during the late stance phase, but did not show significant differences during the early (p=0.238) and mid stance phase (p=0.307). However, there were no significant differences in either the hip or the knee internal rotation moments (p>0.05) nor in the hip adduction angle and moments (p>0.05).

Power calculation:

Hip internal rotation angle was the primary outcome of this study, and thus a post hoc power calculation with G-Power (Version 3.1.9.2) (n=22, two tailed t-test) was performed on this measure for the entire stance phase. The calculated effect size was Cohen's d= 0.57 (Cohen's dz= 0.69) and thus a power of 87% was reached.

Discussion

To the authors' knowledge this is the first study that has investigated hip and knee kinematics and kinetics during running with and without a strap that is designed to decrease hip internal rotation and thereby modify lower limb alignment. This study showed that the Powers[TM] strap significantly reduced the hip and knee internal rotation angle throughout the stance phase. The

strap was effective in correcting an internal rotation of the hip towards a neutral alignment in the transverse plane of the hip. Thus, the first null hypothesis suggesting that the Powers[TM] strap would not modify the hip internal rotation angle was rejected. This is important from a mechanistic perspective as even in individuals who do not have pain, internal rotation can be reduced with the strap and gives confidence that this change was not influenced by pain. The strap also modified the knee internal rotation towards a neutral transverse alignment; however the changes were lower than those of the hip internal rotation. Thus, the second null hypothesis suggesting no changes in the kinetics and kinematics of the hip and knee joint with the Powers[TM] strap could only partially be rejected. Because despite the kinematic changes of the transverse plane of the knee joint no other changes could be observed with the Powers[TM] strap.

The assumption that a transverse correction of the hip might decrease the dynamic knee valgus could not be confirmed in this study because the knee and hip adduction kinematics and kinetics were not significantly modified. One reason for this might be because the participants did not show an excessive dynamic knee valgus during running and their hip and knee adduction angles were in the normal range of motion compared to previous studies [26, 28].

To date, only limited research on knee braces, straps and patellar taping is available, with heterogeneous findings [12, 13, 29]. Studies that investigated the influence of knee braces, straps and patellar taping in patients with running related injuries, such as after an ACL reconstruction or in patients with patellofemoral pain, concluded that bracing or taping does not seem to help function and stability [13, 14, 29]. Studies that analysed the use of knee braces in sports to prevent lower limb injuries reported insufficient evidence to confirm that knee braces may prevent lower limb injuries and lead to optimal training loads [30]. Thus, the evidence that patellar taping and knee braces could modify lower limb biomechanics in patients with running related injuries is still lacking [12-14, 30, 31]. One reason for the lacking evidence is that current research shows a great heterogeneity in the types and use of knee braces, straps and taping techniques.

This study showed that the Powers[TM] strap has the potential to decrease excessive hip internal rotation. This could be due to the Powers[TM] strap being fundamentally different from most knee braces, straps and sleeves that aim to provide a local stabilisation of the knee and the patella. The Powers[TM] strap aims to decrease an excessive internal rotation of the hip, which is associated

with running related injuries, such as patellofemoral pain or ACL ruptures [6]. A reduction of the hip internal rotation during running has to date not been achieved with any other treatment approaches, such as running retraining, straps, braces or patellar tapes. Thus, the Powers[TM] strap might be a promising treatment approach to treat patients with a symptomatic excessive hip internal rotation.

Similar to any other studies, there were some limitations in regards to the findings of the study. It is important to note that the Powers[TM] strap was tested and assessed in the healthy participants where no abnormal range of motion was expected or identified. However, the biomechanical concept of the strap was being tested and thus using healthy participants was the first step in determining its biomechanical effectiveness. In addition, a post-hoc power calculation was carried out for the hip internal rotation angle during stance phase, which revealed a medium effect size and a power of 87%. Thus, it could be concluded that the results presented are significant. However, the reduction of the hip and knee internal rotation were ranging from 1.6° to 4.9° and although these changes were statistically significant, they might not be clinically significant for patients with running related injuries, such as patellofemoral pain.

Since the study focussed on the influence of the Powers[TM] strap on the hip internal rotation, only the influence on the hip and knee biomechanics was analysed. The study showed that the strap significantly influenced the knee kinematics and thus, it is unknown if there were changes in foot biomechanics and this should be investigated in future studies.

Furthermore, 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.

Conclusion

In conclusion, this study has demonstrated that the Powers[TM] strap could alter the transverse plane rotations of the hip and knee and might be a therapy to prevent excessive internal rotation of the hip. Future research should investigate the influence of the Powers[TM] strap on the lower

limb kinematics and kinetics in subjects who show an excessive hip internal rotation during running, such as patellofemoral pain.

References:

[1] Rauh MJ, Koepsell TD, Rivara FP, Margherita AJ, Rice SG. Epidemiology of musculoskeletal injuries among high school cross-country runners. Am J Epidemiol. 2006;163:151-9.

[2] Wirtz AD, Willson JD, Kernozek TW, Hong DA. Patellofemoral joint stress during running in females with and without patellofemoral pain. Knee. 2012;19:703-8.

[3] Taunton JE, Ryan MB, Clement DB, McKenzie DC, Lloyd-Smith DR, Zumbo BD. A retrospective case-control analysis of 2002 running injuries. Br J Sports Med. 2002;36:95-101.

[4] Taylor-Haas JA, Hugentobler JA, DiCesare CA, Hickey Lucas KC, Bates NA, Myer GD, et al. Reduced hip strength is associated with increased hip motion during running in young adult and adolescent male long-distance runners. Int J Sports Phys Ther. 2014;9:456-67.

[5] Schmitz A, Russo K, Edwards L, Noehren B. Do novice runners have weak hips and bad running form? Gait Posture. 2014;40:82-6.

[6] Powers CM. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. J Orthop Sports Phys Ther. 2010;40:42-51.

[7] Nakagawa TH, Serrao FV, Maciel CD, Powers CM. Hip and knee kinematics are associated with pain and self-reported functional status in males and females with patellofemoral pain. Int J Sports Med. 2013;34:997-1002.

[8] Noehren B, Pohl MB, Sanchez Z, Cunningham T, Lattermann C. Proximal and distal kinematics in female runners with patellofemoral pain. Clin Biomech (Bristol, Avon). 2012;27:366-71.

[9] Noehren B, Scholz J, Davis I. The effect of real-time gait retraining on hip kinematics, pain and function in subjects with patellofemoral pain syndrome. Br J Sports Med. 2011;45:691-6.

[10] Willy RW, Scholz JP, Davis IS. Mirror gait retraining for the treatment of patellofemoral pain in female runners. Clin Biomech (Bristol, Avon). 2012;27:1045-51.

[11] Hunt MA, Simic M, Hinman RS, Bennell KL, Wrigley TV. Feasibility of a gait retraining strategy for reducing knee joint loading: increased trunk lean guided by real-time biofeedback. J Biomech. 2011;44:943-7.

[12] Yeung SS, Yeung EW, Gillespie LD. Interventions for preventing lower limb soft-tissue running injuries. Cochrane Database Syst Rev. 2011:CD001256.

[13] Smith TO, Drew BT, Meek TH, Clark AB. Knee orthoses for treating patellofemoral pain syndrome. Cochrane Database Syst Rev. 2015:CD010513.

[14] Rodriguez-Merchan EC. Knee Bracing After Anterior Cruciate Ligament Reconstruction. Orthopedics. 2016;39:e602-9.

[15] Herrington L. Effect of a SERF strap on pain and knee-valgus angle during unilateral squat and step landing in patellofemoral patients. J Sport Rehabil. 2013;22:27-32.

[16] Myer GD, Ford KR, Di Stasi SL, Foss KD, Micheli LJ, Hewett TE. High knee abduction moments are common risk factors for patellofemoral pain (PFP) and anterior cruciate ligament (ACL) injury in girls: is PFP itself a predictor for subsequent ACL injury? Br J Sports Med. 2015;49:118-22.

[17] Hart HF, Culvenor AG, Collins NJ, Ackland DC, Cowan SM, Machotka Z, et al. Knee kinematics and joint moments during gait following anterior cruciate ligament reconstruction: a systematic review and meta-analysis. Br J Sports Med. 2016;50:597-612.

[18] Aminaka N, Pietrosimone BG, Armstrong CW, Meszaros A, Gribble PA. Patellofemoral pain syndrome alters neuromuscular control and kinetics during stair ambulation. J Electromyogr Kinesiol. 2011/04/29 ed2011. p. 645-51.

[19] Dos Reis AC, Correa JC, Bley AS, Rabelo ND, Fukuda TY, Lucareli PR. Kinematic and Kinetic Analysis of the Single-Leg Triple Hop Test in Women With and Without Patellofemoral Pain. J Orthop Sports Phys Ther. 2015:1-26.

[20] Callaghan MJ, Selfe J, McHenry A, Oldham JA. Effects of patellar taping on knee joint proprioception in patients with patellofemoral pain syndrome. Man Ther. 2008;13:192-9.

[21] Birmingham TB, Bryant DM, Giffin JR, Litchfield RB, Kramer JF, Donner A, et al. A randomized controlled trial comparing the effectiveness of functional knee brace and neoprene sleeve use after anterior cruciate ligament reconstruction. Am J Sports Med. 2008;36:648-55.

[22] Cappozzo A, Catani F, Croce UD, Leardini A. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin Biomech (Bristol, Avon). 1995;10:171-8.

[23] Bell AL, Pedersen DR, Brand RA. A Comparison of the Accuracy of Several Hip Center Location Prediction Methods. J Biomech. 1990;23:617-21.

[24] Kristianslund E, Krosshaug T, van den Bogert AJ. Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention. J Biomech. 2012;45:666-71.

[25] Davis RB, Ounpuu S, Tyburski D, Gage JR. A Gait Analysis Data-Collection and Reduction Technique. Hum Movement Sci. 1991;10:575-87.

[26] Alenezi F, Herrington L, Jones P, Jones R. How reliable are lower limb biomechanical variables during running and cutting tasks. J Electromyogr Kinesiol. 2016;30:137-42.

[27] Perry J, Burnfield JM. Gait Analysis: Normal and Pathological Function. Thorofare, USA: Slack Incorporated; 2010.

[28] Novacheck TF. The biomechanics of running. Gait Posture. 1998;7:77-95.

[29] Crossley KM, van Middelkoop M, Callaghan MJ, Collins NJ, Rathleff MS, Barton CJ. 2016 Patellofemoral pain consensus statement from the 4th International Patellofemoral Pain Research Retreat, Manchester. Part 2: recommended physical interventions (exercise, taping, bracing, foot orthoses and combined interventions). Br J Sports Med. 2016.

[30] Salata MJ, Gibbs AE, Sekiya JK. The effectiveness of prophylactic knee bracing in american football: a systematic review. Sports Health. 2010;2:375-9.

[31] Pietrosimone BG, Grindstaff TL, Linens SW, Uczekaj E, Hertel J. A systematic review of prophylactic braces in the prevention of knee ligament injuries in collegiate football players. J Athl Train. 2008;43:409-15.

Table 1: The lower extremity kinematics during stance phase

The kinematic variables (º) during stance phase		Without strap[a]	With strap[a]	95% Confidence Interval[c]		Std. Error Mean[d]	t-test, sig (2-tailed)
				Lower	Upper		
Early stance phase	Hip flexion angle	36.7± 7.4	35.4± 7.0	0.1	2.7	0.8	0.071
	Hip adduction angle	8.3± 3.7	8.2± 3.7	-0.7	0.9	0.4	0.842
	Hip internal rotation angle	4.3± 7.0	1.1± 8.3	0.8	5.5	1.1	0.011[b]
	Knee flexion angle	29.6± 4.8	29.2± 4.7	-1.2	1.9	0.8	0.626
	Knee adduction angle	2.6± 3.6	2.1± 4.0	-0.3	1.2	0.4	0.238
	Knee internal rotation angle	-1.4± 5.1	-3.0± 6.0	0.2	3.1	0.7	0.025[b]
Mid stance phase	Hip flexion angle	37.4± 8.5	36.0± 7.3	-0.4	3.1	0.9	0.116
	Hip adduction angle	11.4± 3.9	10.9± 4.5	-0.7	1.7	0.6	0.374
	Hip internal rotation angle	4.2± 6.6	0.8± 6.8	1.6	5.2	0.9	0.001[b]
	Knee flexion angle	42.0± 4.9	41.6± 4.5	-0.6	1.4	0.5	0.361
	Knee adduction angle	3.6± 3.2	3.2± 3.9	-0.4	1.2	0.4	0.307
	Knee internal rotation angle	3.9± 6.5	1.9± 7.2	0.8	3.1	0.5	0.002[b]
Late stance phase	Hip flexion angle	5.1± 5.2	5.3± 6.8	-2.8	2.5	1.3	0.895
	Hip adduction angle	-0.3± 3.0	-0.8± 3.6	-0.7	1.7	0.6	0.371
	Hip internal rotation angle	3.8± 6.9	-1.1± 7.3	2.4	7.3	1.2	0.001[b]
	Knee flexion angle	23.0± 4.4	24.1±5.8	-3.3	1.1	1.1	0.321
	Knee adduction angle	2.0± 2.6	1.1± 3.0	0.1	1.6	0.4	0.034[b]
	Knee internal rotation angle	-7.4± 7.1	-7.4±7.9	-1.8	1.8	0.9	0.985

[a]Mean ± standard deviation (SD), [b]Significant (P < .05), [c]95% Confidence Interval of the difference, [d] estimated SD of the sample mean

Table 2: The lower extremity kinetics during stance phase

The kinetic variables (Nm/kg) during stance phase		Without strap[a]	With strap[a]	95% Confidence Interval[c]		Std. Error Mean[d]	t-test, sig (2-tailed)
				Lower	Upper		
Early stance phase	Hip flexion moment	1.75± 0.58	1.66± 0.45	-0.09	0.11	0.05	0.469
	Hip adduction moment	1.47± 0.31	1.30± 0.38	-0.04	0.24	0.07	0.135
	Hip internal rotation moment	0.03± 0.23	0.05± 0.13	-0.12	-0.01	0.03	0.556
	Knee flexion moment	1.38± 0.39	1.36± 0.44	-0.11	0.25	0.09	0.774
	Knee adduction moment	0.61± 0.23	0.59± 0.25	-0.04	0.13	0.04	0.607
	Knee internal rotation moment	0.26± 0.11	0.26± 0.09	-0.03	0.03	0.01	0.889
Mid stance phase	Hip flexion moment	1.27± 0.66	1.19± 0.57	-0.11	0.15	0.06	0.429
	Hip adduction moment	2.00± 0.24	1.81± 0.66	-0.16	0.50	0.16	0.240
	Hip internal rotation moment	-0.01± 0.06	0.01± 0.15	-0.05	0.05	0.02	0.744
	Knee flexion moment	2.57± 0.45	2.50± 0.54	-0.14	0.19	0.08	0.480
	Knee adduction moment	0.85± 0.32	0.75± 0.42	-0.10	0.28	0.09	0.290
	Knee internal rotation moment	0.46± 0.13	0.44± 0.12	-0.03	0.03	0.01	0.418
Late stance phase	Hip flexion moment	-0.07± 0.35	-0.12± 0.28	-0.04	0.14	0.04	0.380
	Hip adduction moment	0.24± 0.14	0.23± 0.14	-0.11	0.00	0.03	0.772
	Hip internal rotation moment	0.01± 0.03	0.02± 0.01	-0.03	0.00	0.01	0.157
	Knee flexion moment	-0.04± 0.14	0.04± 0.16	-0.09	-0.01	0.02	0.063
	Knee adduction moment	0.09± 0.11	0.09± 0.12	-0.05	0.01	0.01	0.822
	Knee internal rotation moment	0.02± 0.04	0.02± 0.03	-0.01	0.01	0.00	0.180

[a]Mean ± standard deviation (SD), [b]Significant (P < .05), [c]95% Confidence Interval of the difference, [d] estimated SD of the sample mean

Figure 1.: The application of the markers and the Powers^{TM} strap (The Left medial knee marker is not shown. A the static reference trial was collected without the applied Powers^{TM} strap for both conditions.)
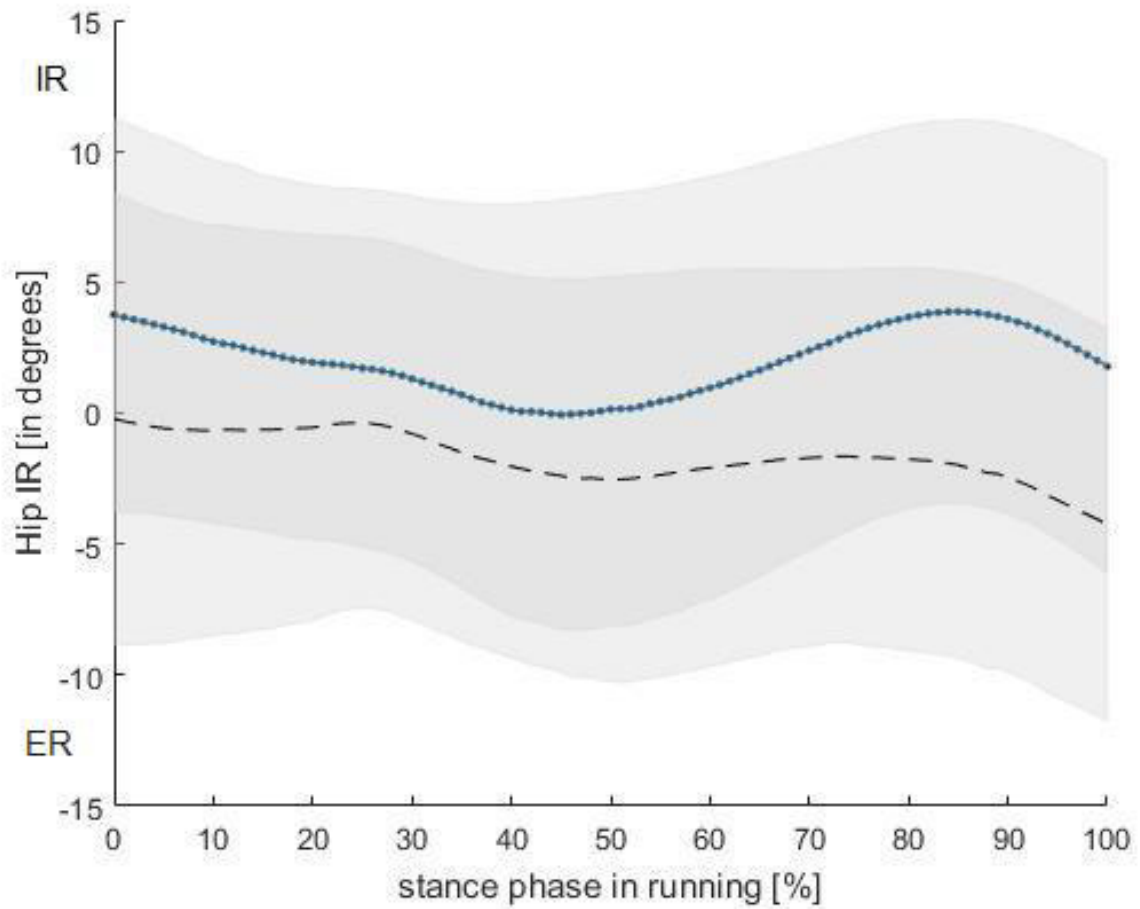
Figure 2: The transverse plane hip angle during during the stance phase of running under 2 conditions: without (dotted line) and with the Powers^{TM} strap (dashed line). The shaded areas represent ±1SD for each condition, internal rotation as the positive angle.
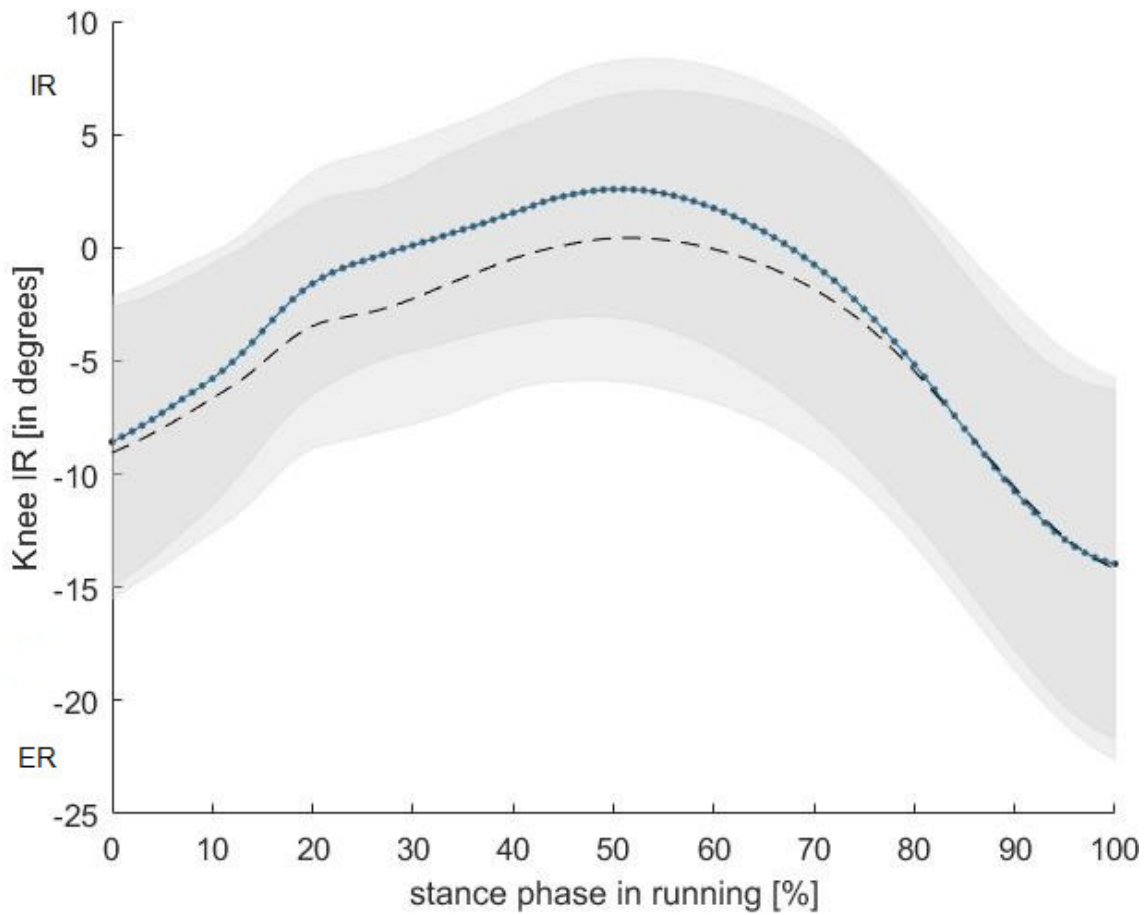
Figure 3: The transverse plane knee angle during the stance phase of running under 2 conditions: without (dotted line) and with the Powers^{TM} strap (dashed line). The shaded areas represent ±1SD for each condition, internal rotation as the positive angle.

Highlights:

1. The Powers™ strap decreased hip internal rotation during the stance phase in running.

2. The Powers™ strap did not modify hip or knee joint kinetics during the stance phase in running.

3. The Powers™ strap might be a promising treatment approach to treat patients with an excessive hip internal rotation.