Osteoarthritis and Cartilage 20 (2012) 1500–1506

Osteoarthritis and Cartilage

The effect of walking poles on the knee adduction moment in patients with varus gonarthrosis

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ARTICLE INFO

Article history:
Received 25 March 2012
Accepted 22 August 2012

Keywords:
Osteoarthritis
Knee joint load
Gait
Assistive walking devices
Hiking poles

SUMMARY

Objectives: (1) Test the hypothesis that walking poles decrease the external knee adduction moment during gait in patients with varus gonarthrosis, and (2) explore potential mechanisms.
Design: Thirty-four patients with medial compartment knee osteoarthritis (OA) and varus alignment underwent three dimensional (3D) gait analysis with and without using walking poles. Conditions were randomized and walking speed was maintained ±5% of the self-selected speed of the initial condition. The pole held in the hand of the unaffected side was instrumented with a compression load cell.
Results: Student's t tests for paired samples indicated small but statistically significant increases ($P < 0.001$) in knee adduction moment (calculated from inverse dynamics) for its first peak, second peak and angular impulse when using the poles; mean increases (95% confidence interval – CI) were 0.17% BW*Ht (0.08, 0.27), 0.17%BW*Ht (0.04, 0.30) and 0.15%BW*Ht*s (0.09, 0.22), respectively. There was a decrease ($P = 0.015$) in vertical ground reaction force (−0.02 BW (−0.04, −0.01)), yet increase ($P < 0.001$) in its frontal plane lever arm about the knee (0.30 cm (0.15, 0.44)), at the time of the first peak knee adduction moment. Pole force in the vertical direction was inversely related ($r = -0.34$, $P = 0.05$) to the increase in first peak adduction moment.
Conclusion: Although results are variable among patients, and may be related to individual technique, these overall findings suggest that walking poles do not decrease knee adduction moments, and therefore likely do not decrease medial compartment loads, in patients with varus gonarthrosis. Decreases in knee joint loading should not be used as rationale for walking pole use in these patients.

Introduction

Conservative strategies are recommended as first line treatments for knee osteoarthritis (OA). These include patient education, exercise and attempts to lessen knee joint loads[1,2]. Nordic walking poles, also called hiking poles, may be appealing to patients with knee OA because the poles are promoted by manufactures as a method to reduce the stress on the joints of the lower extremities while enabling increased fitness (urbanpoling.com, 2011). Walking poles differ from other assistive devices for patients with knee OA such as canes. Canes are typically used singly, in one hand, and are primarily intended to decrease pain by lessening the load on the hip and knee. Alternatively, walking poles are typically used in pairs, in an alternating pattern, and are suggested to help propel the user forward[3]. While canes typically decrease walking speed[4], walking poles increase walking speed[3].

Most recent paradigms suggest that knee OA involves local biomechanical factors acting within the context of systemic factors, with varus alignment of the lower limb being particularly important[5–9]. Biomechanical evidence from a variety of sources demonstrates that the load borne by the medial compartment of the tibiofemoral joint is substantially higher than the lateral compartment – a phenomenon observed in healthy joints and in neutral alignment, yet exacerbated markedly by varus alignment[10–12]. Consistent with the importance of frontal plane alignment, gait studies suggest that the external knee adduction moment is a valid[13–17] and reliable[18] proxy for the dynamic load on the medial compartment in patients with knee OA. High knee adduction moments have also been shown to predict the future onset of knee pain[19] and radiographic[20] and MRI measures[21] of disease progression in the medial compartment. Consequently,

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http://dx.doi.org/10.1016/j.joca.2012.08.014

although limitations do exist and must be considered carefully[22], the efficacy of several interventions, such as unloader knee braces[23], foot orthotics[24] and gait retraining[25] has been tested by evaluating their effects on the knee adduction moment. The knee adduction moment is calculated through inverse dynamics using three dimensional (3D) gait kinematics and kinetics, and is most largely influenced by its frontal plane lever arm and ground reaction force. Accordingly, information about the mechanisms underlying changes in the knee adduction moment can be gained by also investigating those measures[5,26–28].

Results from previous studies evaluating the effect of walking poles on the knee adduction moment vary considerably[22,29,30]. Stief *et al.*[30] reported an increase in the first peak knee adduction moment with walking poles in healthy, experienced walking pole users. Walter *et al.*[22] reported decreases in the first and second peaks of the knee adduction moment with walking poles for a subject with an instrumented total knee replacement. Jensen *et al.*[29] reported no change in knee adduction moment with walking poles used by healthy subjects applying a range of loads to the poles. Importantly, patients with medial compartment knee OA and varus alignment (i.e., varus gonarthrosis) typically have greater knee adduction moments than healthy adults or patients after total knee replacement, and therefore may respond differently to walking pole use.

Inconsistencies in previous findings may be partly due to variability in how participants use walking poles[22,29,30]. Specifically, some people may use the poles much like a cane. They may direct substantial force downward through the pole (held in the hand opposite to the affected limb) and create a frontal plane moment about the knee that tends to oppose the knee adduction moment, whereas other participants may not[4,25,31]. Therefore, when evaluating the effect of walking poles on the knee adduction moment in patients with knee OA, measuring the force on the pole in the frontal plane and angle at which the pole is held in the sagittal plane may help explain any observed changes.

We are unaware of any previous investigations that have evaluated the effects of walking poles on the knee adduction moment in patients with knee OA. The objectives of this study were to: (1) test the hypothesis that walking poles decrease the external knee adduction moment during gait in patients with varus gonarthrosis, and (2) explore potential mechanisms for decreasing the adduction moment by evaluating gait variables likely to influence it most.

Methods

Participants

Thirty-four patients (22 male) with no experience using walking poles were recruited from a tertiary care clinic specializing in orthopaedics. Inclusion criteria were: pain in one knee with radiographically confirmed varus gonarthrosis, defined as varus alignment (mechanical axis angle $\leq -1°$) and OA of greatest severity in the medial compartment of the tibiofemoral joint. Diagnosis of OA was based on Altman's criteria[32]. One trained investigator (DJB) measured the mechanical and anatomical axis angles from standing anteroposterior hip-to-ankle radiographs using custom software[33]. Severity of OA was graded by two investigators (DB and KML) using the Kellgren and Lawrence scale[34]. In the event of any discrepancies, the X-rays were re-graded and a consensus was reached. Exclusion criteria included inflammatory or infectious arthritis of the knee, other musculoskeletal or neurological conditions likely to affect gait, unable to speak/read English, or inability to understand and provide informed consent. Sample size was based on the ability to detect a significant ($P < 0.05$) within-subject difference in the knee adduction moment between conditions (with poles and without poles) of medium effect size ($d = 0.5$) 80% of the time (G*Power Version 3.1.1, Universitat Kiel, Germany). This study was approved by the institution's Research Ethics Board for Health Sciences involving human participants. All patients provided informed, signed consent.

Gait analysis

Patients visited the laboratory on two occasions. The first visit consisted of a 30 min introductory session delivered by a trained walking pole instructor. During this session, patients were given an overview of the study, walking poles, and technique. Patients were instructed on adjusting pole length and walking technique based on the manufacturer's recommendations (urbanpoling.com, 2011). Recommendations included: (1) maintaining trunk in a 'tall' upright position, (2) walking with contralateral pole and heel contacting the ground simultaneously, (3) contacting the tips of the poles with the ground at a point just behind the contralateral heel, (4) lifting the handle of the pole to a 'handshake position' as the ground is contacted, and (5) pressing down on the handle with a comfortable grip and extended elbows. Patients completed several practice trials under the guidance of the instructor until s/he felt comfortable using the poles with the described technique. After the training session was completed, patients were given access to a web-based instructional video and sent home with a set of poles for a period of 1 week. Patients were instructed to practice using the poles during prolonged walking and to record the number of days where the poles were used for at least 20 min.

During the second visit, patients underwent 3D gait analysis using an eight-camera motion capture system (Eagle HiRes cameras, EvaRT 4.2 system, Motion Analysis Corp., Santa Rosa, CA, USA) synchronized with a floor-mounted force plate (OR6 model, Advanced Mechanical Technology, Watertown, MA, USA). The patients wore comfortable shorts and a t-shirt.

A modified Helen Hayes marker set consisting of 22 passive reflective markers was used[35]. An additional marker was placed on the right scapula to aid the identification of sides during processing. Patients first stood motionless on the force plate to determine body mass, relative marker orientations, and the locations of the knee and ankle joint centres. Additional markers were placed on the medial aspect of the knees and medial malleoli for this static trial and were removed prior to gait testing. Patients then walked barefoot across an 8 m walkway at a self-selected speed. The velocity of the sacral marker in the forward direction was monitored during all trials and patients were provided feedback to alter gait speed if changes greater than 5% were observed. Only trials with walking speeds within 5% of the first condition were processed. Patients performed walking trials until five complete force plate foot strikes of the affected limb were collected for each condition. The conditions of walking with or without poles were randomized using a random number generator conditioned for 0 and 1. Zero represented 'with poles' being performed first ($n = 16$). A one meant the 'without poles' condition was performed first ($n = 18$). Pain intensity after each condition was assessed using an eleven point numerical rating scale. Patients were asked to rate the pain in their affected knee at the beginning of the visit and at the completion of each condition. Zero indicated no pain and 10 indicated worst pain imaginable.

Primary outcome measure

The external adduction moment about the knee was calculated from patient anthropometrics, kinematic (sampled at 60 Hz) and kinetic data (sampled at 1,200 Hz) using inverse dynamics (Orthotrak 6.2.4; Motion Analysis Corporation, Santa Rosa, CA)[18].

Kinematic data were filtered using a no-lag fourth order Low-Pass Butterworth filter with a 6 Hz cut off frequency. Kinetic data were filtered using a 1,000 Hz two-pole, low-pass filter. Each lower limb segment (foot, shank, and thigh) was modelled as a rigid body with a local coordinate system that coincided with anatomical axes. Inertial properties of each limb segment were approximated anthropometrically and translations and rotations of each segment were reported relative to neutral positions defined during the initial standing static trial. For each trial, the knee adduction moment waveform was plotted over 100% of stance, normalized to body weight and height (%BW*Ht), and inspected visually. The peak magnitudes of the external knee adduction moment in the first and second halves of stance were identified using an algorithm that identified values immediately preceded by a minimum of five continuously ascending values and immediately followed by a minimum of five continuously descending values. If no identifiable peak occurred in a given half of stance, no knee adduction moment value for that half of stance was recorded. The entire knee adduction moment waveform (not expressed as a percentage of stance) was also summarized as its angular impulse (i.e., the area under the curve in %BW*Ht*s).

Secondary outcome measures

To investigate the most likely mechanisms for changing the knee adduction moment with walking poles, vertical ground reaction force, frontal plane lever arm, gait speed and lateral trunk lean towards the stance limb at the time of first and second peak knee adduction moment were also identified[26,36]. The frontal plane lever arm was calculated as the perpendicular distance between the resultant frontal plane ground reaction force line of action and the centre of rotation of the knee[26]. Gait speed was calculated based on the average speed of the sacral marker during successive foot strikes of the affected limb. The maximum lateral trunk lean angle was calculated as the angle of a line drawn from the midpoint of the anterior superior iliac spines to the midpoint of the anterior tips of the acromion processes with respect to the vertical[36]. Positive values indicated a trunk lean towards the stance limb, while a negative value indicated a lean towards the swing limb. Given the suggested importance of considering the external knee flexion moment in combination with the adduction moment when evaluating potential changes in knee joint loads with gait modifications[22], we also evaluated the first peak (flexion) and second peak (extension) sagittal plane knee moment during stance.

Additional markers were placed at the base of the handle and at the distal tip of the walking poles. The walking pole (Urban Poling Inc., BC, Canada) carried in the contralateral hand to the affected knee was instrumented with a compression load cell (Model LC201-300; Omegadyne Inc., QC, Canada) and telemetry system attached near the handle of the pole. The pole was cut into two pieces directly above the inferior end. One end of the pole was fastened to each side of the load cell. The load cell was calibrated using the laboratory force plate by aligning the pole with the vertical axis of the laboratory and applying different magnitudes of force. Load cell and force plate data were synchronized and simultaneously recorded (EvaRT 4.2; Motion Analysis Corp., Santa Rosa, USA).

Pole force in the vertical direction was determined by separating the measured axial pole force into its orthogonal components assuming that pole force acted exclusively along its long axis. Sagittal plane pole angle was defined as the angle of the line drawn from the inferior to the superior pole marker with respect to the laboratory's horizontal axis (0°) in the direction of travel[31]. The pole force in the vertical direction, and the sagittal plane pole angle, were selected because we hypothesized they had the most potential to vary among patients and correlate to the knee adduction moment.

Statistical analysis

Variables were averaged over five trials for each condition for each patient. Means and standard deviations (SDs) during walking with and without poles, and mean changes with 95% confidence intervals (CIs), were then calculated for all measures. Normality was tested with Shapiro–Wilks tests. For the primary objective, changes in the external knee adduction moment measures were compared using student's *t* tests for paired samples. A two-sided P value of 0.05 was used to denote statistical significance. Statistical analysis was performed using the Statistical Package for the Social Sciences (SPSS v. 20; SPSS Inc., Chicago, IL).

For the second objective, secondary outcomes were also compared with and without walking poles using student's *t* tests for paired samples. The relationships between the change in peak knee adduction moment, the pole force in the vertical direction and the sagittal plane pole angle were then explored using Pearson correlation coefficients.

Results

Patient demographics are presented in Table I. Patients reported using the poles a mean (SD) of 3.2 (1.6) days in the week between initial instruction and gait analysis. Walking speed was 1.17 (0.18) m/s with poles and 1.16 (0.19) m/s without poles ($P = 0.45$). Patients reported an average pain intensity of 1.5 (1.5) at the start of the session, 1.9 (1.8) at the end of the condition with poles, and 1.8 (1.7) at the end of the condition without poles ($P = 0.23$). Pain while walking with the poles decreased for four patients, increased for seven, and remained unchanged for the remaining 23.

Variables observed with and without poles, including their difference scores were normally distributed. There were small but statistically significant ($P \leq 0.001$) increases, rather than deceases, in the knee adduction moment first peak, second peak and angular impulse when using the poles. Descriptive statistics for all variables measured at the first and second peaks of the knee adduction moment are presented in Tables II and III, respectively. The mean (SD) knee adduction angular impulse was 1.53 (0.46) %BW*Ht*s with the poles and 1.38 (0.42) %BW*Ht*s without the poles. Ensemble average waveforms ($n = 34$) for the knee adduction moment, frontal plane lever arm and vertical ground reaction force throughout stance during both conditions are illustrated in Fig. 1. There was an increase in lever arm length ($P < 0.001$), a decrease in vertical ground reaction force ($P = 0.015$) and a decrease in trunk lean ($P < 0.001$) at the time of the first peak knee adduction

Table I
Patient demographic and clinical characteristics

Age (years)	53.6 (9.8)
Height (m)	1.74 (0.10)
Mass (kg)	88.1 (16.0)
BMI (kg/m^2)	28.3 (5.4)
Number of males	22
Number of females	12
Kellgren and Lawrence grade (number of patients)	
	Medial
0 or 1	0
2	18
3	11
4	4
Mechanical axis angle (degrees)*	−6.5 (2.8)
Anatomical axis angle (degrees)*	−1.3 (2.8)

Values are means (SD) unless stated otherwise.
[*] Negative values indicate varus alignment.

Table II
Kinetic and kinematic measures at the time of first peak knee adduction moment with and without walking poles

	With poles mean (SD)	Without poles mean (SD)	Mean difference (95% CI)
Primary outcome measure			
First peak knee adduction moment (%BW*Ht)	2.88 (0.79)	2.71 (0.78)	0.17 (0.08, 0.27)[*]
Secondary outcome measures			
Frontal plane lever arm (cm)	5.27 (1.45)	4.97 (1.35)	0.30 (0.15, 0.44)[*]
Lateral trunk lean (deg)	0.12 (1.70)	1.33 (1.65)	−1.21 (−1.59, −0.84)[*]
Vertical ground reaction force (BW)	0.99 (0.09)	1.02 (0.08)	−0.02 (−0.04, −0.01)[*]
Peak knee flexion moment (%BW*Ht)	1.07 (1.64)	1.00 (1.41)	0.07 (−0.14, 0.29)
Axial pole force (N)[†]	44.0 (26.2)		
Vertical pole force (N)[†]	39.3 (24.5)		
Sagittal pole angle (deg)	64.2 (11.7)		

[*] Indicates significant difference between conditions $P < 0.05$.
[†] Axial pole force was the force measured by the load cell, whereas vertical pole force was the portion of pole force applied in the vertical direction.

moment when using the poles. There was no significant difference in frontal plane lever arm ($P = 0.88$) or vertical ground reaction force ($P = 0.19$), and a small but statistically significant decrease in lateral trunk lean ($P = 0.001$), at the time of the second peak knee adduction moment when using the poles.

The correlation between the change in the first peak knee adduction moment and the force applied through the pole in the vertical direction (Fig. 2), and the pole angle in the sagittal plane were, $r = -0.34$, $P = 0.05$ and $r = -0.25$, $P = 0.16$, respectively. The correlation between the change in the second peak knee adduction moment and the force applied through the pole in the vertical direction, and the pole angle in the sagittal plane, were $r = 0.01$, $P = 0.96$ and $r = 0.03$, $P = 0.86$. Mean peak axial pole force was 63.8 (36.5)N and occurred at 25.9 (14.0)% stance. The correlation between the change in the first peak knee adduction moment and the peak pole force was $r = -0.37$, $P = 0.03$.

There was no significant difference in the peak flexion moment between conditions (Table II). There was a small but statistically significant ($P < 0.05$) decrease in peak knee extension moment (Table III).

Discussion

Table III
Kinetic and kinematic measures at the time of second peak knee adduction moment with and without walking poles

	With poles mean (SD)	Without poles mean (SD)	Mean difference (95% CI)
Primary outcome measure			
Second peak knee adduction moment (%BW*Ht)	3.05 (0.85)	2.88 (0.82)	0.17 (0.04, 0.30)[*]
Secondary outcome measures			
Frontal plane lever arm (cm)	4.54 (1.63)	4.56 (1.39)	−0.03 (−0.39, 0.34)
Lateral trunk lean (deg)	0.26 (1.64)	0.99 (1.55)	−0.75 (−1.16, −0.32)[*]
Vertical ground reaction force (BW)	0.99 (0.19)	1.03 (0.09)	−0.04 (−0.09, 0.02)
Peak knee extension moment (%BW*Ht)	−2.30 (1.14)	−2.59 (1.04)	0.29 (0.13, 0.45)[*]
Axial pole force (N)[†]	16.1 (22.1)		
Vertical pole force (N)[†]	14.0 (21.0)		
Sagittal pole angle (deg)	52.9 (11.4)		

[*] Indicates significant difference between conditions $P < 0.05$.
[†] Axial pole force was the force measured by the load cell, whereas vertical pole force was the portion of pole force applied in the vertical direction.

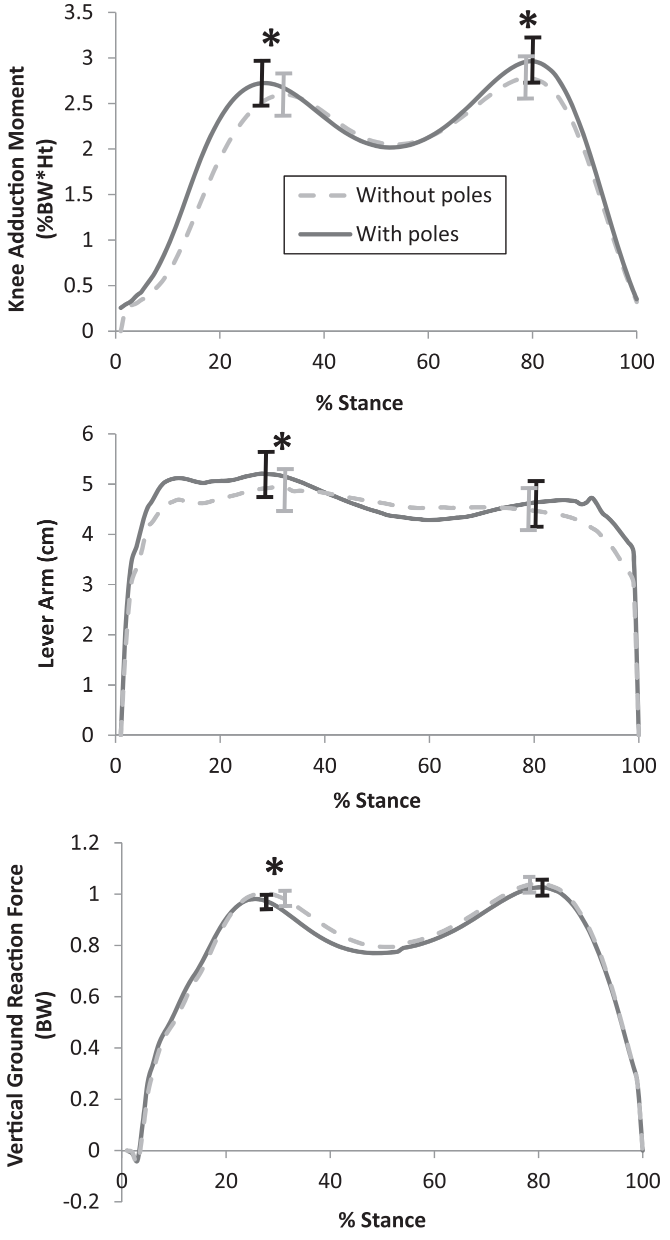

Fig. 1. Ensemble waveform ($n = 34$) of the knee adduction moment, frontal plane lever arm, and vertical ground reaction force over 100 percent stance with and without walking poles. 95% CIs around sample means are shown for all measures at the time of the first and second peak knee adduction moment. *$P < 0.05$ using student's t tests for paired samples.

The present results suggest that walking with poles increases the knee adduction moment in patients with varus gonarthrosis, and therefore do not support the claim that walking poles decrease knee joint loading in these patients. On average, increases in the knee adduction moment first peak, second peak and angular impulse were 6%, 10%, and 12%, respectively. Some authors have previously questioned the clinical importance of changes in the knee adduction moment of this magnitude for reasons such as only moderate correlations with direct medial compartment contact force[37], or limited carry over to pain, performance, or disability[38]. Alternatively, other authors[21,39] have suggested that small increases or decreases in the knee adduction moment could have substantial effects on patients with knee OA given the thousands of steps taken per day[40,41] and the reported influence on disease progression[20,21].

The present results also suggest that the mechanism for increased knee adduction moments when patients with varus gonarthrosis use poles is their effect on the frontal plane lever arm.

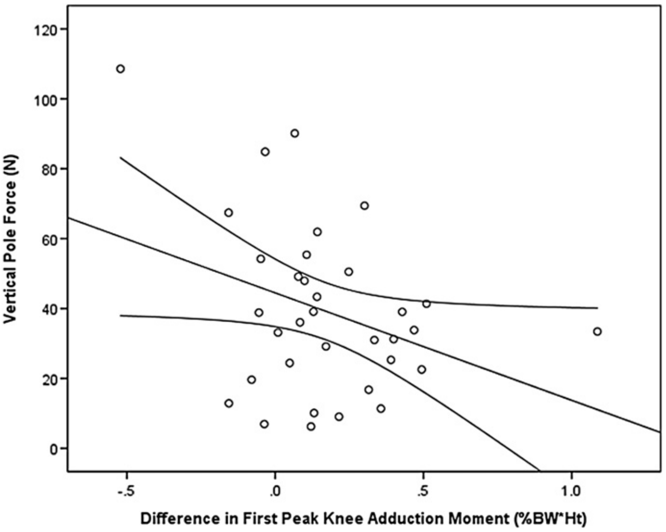

Fig. 2. A scatter plot of the pole force in the vertical direction vs the difference in the first peak knee adduction moment with and without poles. Positive values along the *X* axis indicate an increase in the first peak knee adduction moment with poles. Pearson $r = -0.34$, $P = 0.05$. 95% mean prediction intervals are shown.

Specifically, patients walked with less lateral trunk lean towards the stance limb when using the walking poles. Interestingly, these findings are consistent with the recommended instruction during training to maintain the trunk in a tall upright position, and may suggest that such instructions are not suitable for patients with varus gonarthrosis. Several investigators have reported the importance of lateral trunk lean and subsequent changes in the frontal plane lever arm to the magnitude of the knee adduction moment[25,36,42]. For example, Mundermann *et al.*[42] reported that walking with increased trunk lean as large as 10° decreased the knee adduction moment by an average of 65% for healthy participants. Hunt *et al.*[36] reported that 13% of the variation in the first peak knee adduction moment was explained by lateral trunk lean in patients with knee OA, after controlling for other factors. Additionally, lateral trunk lean towards the stance limb has been suggested to be a possible compensatory mechanism that patients with knee OA adopt to reduce high knee joint loads[43–46]. Therefore, walking poles may increase the knee adduction moment in patients by inhibiting that potential compensation.

It should be noted that the observed changes in the knee adduction moment were quite variable among patients. This is consistent with previous investigations on walking poles. Stief *et al.*[30] observed a 15% increase during the first peak in healthy volunteers. Alternately, Walter *et al.*[22] observed a 33% and 47% decrease in the first and second peak knee adduction moment (with a concomitant decrease in second peak medial contact force only) in a subject with an instrumented total knee replacement. A small number of patients in the present study also experienced potentially substantial decreases, and increases, in the knee adduction moment when using the poles.

We are aware of only two previous studies that have reported axial walking pole force, both evaluating healthy subjects. Schiffer *et al.*[47] investigated energy cost during pole walking. They reported peak axial pole force between 36.5 and 43.3N depending on the type of walking surface. Our findings suggest that on average, patients applied a comparable pole force of 39.3N. Jensen *et al.*[29] investigated the effects of different magnitudes of pole force on knee joint load. They found no change in first or second peak knee adduction moment, even when pole force was increased by a mean of 2.4 times. Although considerably different from walking poles, Simic *et al.*[48] investigated the effects of varying load application on contralateral cane use in patients with knee OA. They reported a reduction in knee adduction moment first peak, second peak, and angular impulse proportionate to the load placed through the cane. The loads placed through the canes were greater than the loads place on the walking poles here, and emphasize what we believe to be an important difference between these devices. Based on their findings, Simic *et al.*[48] emphasize the potential importance of cane placement and the timing of load application.

In the present patients, the observed change in the first peak knee adduction moment when using walking poles was significantly correlated with the amount of force applied through the pole in the vertical direction. However, the magnitude of the correlation was quite low ($r = -0.34$) and was influenced largely by extreme values. Fig. 2 shows that the greatest decrease in the first peak knee adduction moment was experienced by the patient who applied the greatest load (108.6N) in the vertical direction. Conversely, the greatest increase in first peak knee adduction moment occurred in a patient who applied a force of only 37.2N. Future research is necessary to determine if patients with varus gonarthrosis can be trained to decrease the knee adduction moment by altering the force directed in the vertical direction. Presently, Fig. 2 emphasizes that the effect of poles on the knee adduction moment is highly variable, and that the vast majority of these patients experience small increases when the poles are used as instructed, regardless of force.

It is possible that individuals with varus gonarthrosis use the poles differently than others and may experience different effects. Therefore, care should be taken to restrict generalization of these findings to similar patients. Additionally, patients reported using the poles a mean of 3.2 (1.6) days in the week prior to testing. Additional time spent walking with the poles may influence results. It should also be noted that pole walking tends to increase walking speed[3]. We maintained speed ±5% to preserve internal validity. However, increases in walking speed also tend to increase knee joint loading. Therefore, the present increases in knee adduction moment with pole use may actually be underestimates.

It must also be acknowledged that a change in knee adduction moment observed when using the poles may not necessarily correspond with the same result in the medial compartment load[22]. Given that the direct measurement of medial contact force in a subject (male, age = 83, body mass index (BMI) = 23.5, neutral alignment) with an instrumented knee joint replacement was best predicted by a combination of peak external knee adduction and flexion moments, a reasonable strategy when evaluating interventions intended to decrease medial compartment loads is to consider the combined effect of these two knee moments[22]. When doing so in the present study, these mean peak moments in late stance changed in the opposite directions when using the poles (Table III). It is possible that the generally assumed increase in medial compartment load that would accompany an increase in second peak knee adduction moment was negated by a decrease in peak extension moment. Although the importance of varus alignment and the associated knee adduction moment to aberrant medial compartment loads in patients with varus gonarthrosis must be considered, the inclusion of the peak extension moment in the present study may question whether the poles caused an increase in medial compartment load during late stance or not. Regardless, the overall findings considering both moments throughout all of stance do not support the claim that walking poles decrease knee joint loads.

Conclusions

Although results are variable among patients, and may be related to individual technique, these overall findings suggest that walking poles do not decrease knee adduction moments, and therefore likely do not decrease medial compartment loads, in patients with varus gonarthrosis. Decreases in knee joint loading should not be used as rationale for walking pole use in these patients.

Contributions

All authors contributed to the acquisition, analysis and interpretation of data, article's revisions and final approval before submission.

Role of the funding source

This research was undertaken thanks to funding from the Canada Research Chairs Program (TBB), the Canadian Institutes of Health Research (CIHR) and Arthrex Inc. Urban Poling Inc. provided the walking poles. Arthrex Inc. and Urban Poling Inc. had no role in study design, data collection, data analysis, or writing of the manuscript. KML is supported by the Joint Motion Program, a CIHR Strategic Training Initiative in Health Research.

Conflict of interest

No authors had any financial or personal relationships with other people or organizations that could have influenced this study.

Acknowledgements

None.

Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.joca.2012.08.014.

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