The Influence of Arch Supports on Knee Torques Relevant to Knee Osteoarthritis

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ABSTRACT

FRANZ, J. R., J. DICHARRY, P. O. RILEY, K. JACKSON, R. P. WILDER, and D. C. KERRIGAN. The Influence of Arch Supports on Knee Torques Relevant to Knee Osteoarthritis. Med. Sci. Sports Exerc., Vol. 40, No. 5, pp. 913–917, 2008. Purpose: Changes in footwear and foot orthotic devices have been shown to significantly alter knee joint torques thought to be relevant to the progression if not the development of knee osteoarthritis (OA) in the medial tibiofemoral compartment. The purpose of this study was to determine if commonly prescribed arch support cushions promote a medial force bias during gait similar to medial-wedged orthotics, thereby increasing knee varus torque during both walking and running. Methods: Twenty-two healthy, physically active young adults (age, 29.2 ± 5.1 yr) were analyzed at their self-selected walking and running speeds in control shoes with and without arch support cushions. Three-dimensional motion capture data were collected in synchrony with ground reaction force (GRF) data collected from an instrumented treadmill. Peak external knee varus torque during walking and running were calculated through a full inverse dynamic model and compared. Results: Peak knee varus torque was statistically significantly increased by 6% (0.01 ± 0.02 Nm·(kg·m)⁻¹) in late stance during walking and by 4% (0.03 ± 0.03 Nm·(kg·m)⁻¹) during running with the addition of arch support cushions. Conclusions: The addition of material under the medial aspect of the foot by way of a flexible arch support promotes a medial force bias during walking and running, significantly increasing knee varus torque. These findings suggest that discretion be employed with regard to the prescription of commonly available orthotic insoles like arch support cushions. Key Words: BIOMECHANICS, ORTHOTICS, KINETICS, GAIT, JOINT TORQUES, FOOTWEAR

Knee osteoarthritis (OA) develops in approximately 10% of adults over the age of 55 and represents a major cause of pain and physical disability in the elderly (8,18). The possibility that different types of footwear contribute to the progression if not the development of knee OA deserves strong consideration because footwear is a potentially controllable and easily modifiable factor for this prevalent and disabling disease. We (14–16) previously found that typical forms of women’s shoes significantly exaggerate knee external varus torque, which is thought to be relevant to the progression if not the development of knee OA. Specifically, we showed an increase in varus torque during walking in both wide-based and narrow-based high-heeled shoes with an average heel height of 2.8 inches (7.1 cm) (14,16), and in shoes with heels as moderate as 1.5 inches (3.8 cm) (15). This increase implies exaggerated compressive loading on the medial tibiofemoral compartment (20,27,30), the typical site of knee OA (33). Similarly, Shakoor et al. showed that even comfortable walking shoes increase this varus torque by 11.9% as compared with bare-foot walking in both men and woman with medial compartment OA (29).

Considering the evidence that implicates varus torque in the progression if not the development of knee OA (1,9–12,14–17,19,23,29), conservative interventions designed to reduce medial compartment loading are of great interest. Previous research has shown the effectiveness of foot orthotic devices, specifically of lateral shoe wedges, in reducing knee varus torque and in lessening pain in patients with medial compartment knee OA (4,13,17). Conversely, the use of medial wedges has been found to significantly increase this torque (28). Previous research evaluating the impact of foot orthotic devices, hereafter referred to as orthotics, on
knee torques is not limited to walking alone. In a study motivated by running overuse injuries, Mündermann et al. observed that the addition of a full-length 6-mm medially posted orthotic increased knee rotational torque and significantly delayed peak knee varus torque during running (21). The observation that changes in footwear and orthotics significantly affect knee varus torque suggests that alterations in foot and ankle kinetics cause biomechanical compensations at the knee to maintain stability and forward progression during walking and running. Walking is by far the most common daily activity causing repetitive force through the knee’s medial compartment, and running causes a significantly greater magnitude of this repetitive force (20,22). Excessive repetitive loading is believed to be an important etiological factor in the development of OA (1). In fact, normal walking and running biomechanics are such that coronal plane knee torques are varus throughout stance (9,20), potentially explaining why tibiofemoral knee OA begins predominantly in the medial compartment. Animal data also support that repetitive knee varus loading leads to degenerative changes in the medial compartment of the knee (23).

Although custom-molded orthotics have been proposed as an effective treatment of running overuse injuries (5,6,26), the orthotic devices commonly available through retailers typically provide for comfort alone. For example, arch support cushions are widely prescribed by retailers based upon claims that they aid in shock absorption, provide increased stability and support, and maintain proper foot position during walking and running (Spenco Medical Corporation, Waco, TX; Shock Doctor, Inc., Plymouth, MN; Sting Free Technologies Company, Berwyn, PA). However, it is possible that the addition of material under the medial aspect of the foot by way of a flexible arch support could promote a medial bias during gait similar to medial-wedged orthotics, thereby increasing knee varus torque. Therefore, the purpose of this study was to investigate the effects of arch support cushions on knee torques relevant to knee OA. We hypothesized that commonly available arch supports would, like medial-wedged orthotics, increase knee varus torque during both walking and running.

METHODS

A total of 22 healthy runners (10 female) were recruited from the local population (Table 1). The subjects described their running as recreational, running a minimum of 15 miles each week. Subjects had no history of chronic musculoskeletal pathology and were without a running-related injury at the time of testing. The experimental protocol was approved by the UVA institutional review board for health science research, and written informed consent was obtained from each subject before testing. Subjects were asked to walk at their comfortable walking speed and to run at their submaximal pace on an instrumented treadmill under two conditions: in control running shoes with and without arch cushion orthotics, the order of which was randomized. Randomization was done by alternating the conditions according to the order of the subject’s admission to the study. Submaximal running pace was defined using a rate of perceived exertion (RPE) between 6 and 7 on a scale of 10.

Standardized footwear was provided to all subjects. The control shoe used in this study was the New Balance 755 (New Balance Athletic Shoe, Inc., Boston, MA). This model was a neutral shoe with a single-density midsole and semicurved last. The control shoe was chosen for its neutral classification and lack of advanced structural elements designed to alter force progression through the shoe. The orthotic examined in this study was the Spenco full-length arch support cushion illustrated in Figure 1 (Spenco Medical Corporation, Waco, TX). The insole had a full-length thickness of 4 mm, and an uncompressed arch support height of 26 mm with a support material durometer of 28 shore c. Orthotic insoles of this type have been shown to provide arch support and increased shock attenuation during walking and running (32). Prior to all testing, each subject completed a 4-min warm-up period at his or her self-selected walking and running speeds on the treadmill.

Sixteen retro-reflective markers were placed by the same physical therapist over anatomical landmarks of the pelvis

![FIGURE 1—Full-length arch support cushions.](http://www.acsm-msse.org)
and lower extremity. The three dimensional positions of each marker were captured at 250 Hz using a 10-camera Vicon 624 motion analysis system (Vicon Peak, Lake Forest, CA). Synchronized ground reaction force data were captured at 1000 Hz using an AMTI (AMTI, Watertown MA) instrumented treadmill described in detail elsewhere (24,25). The treadmill consists of two side-by-side force platform units (330 mm x 1395 mm) positioned behind a larger unit (663 mm x 2750 mm) providing a continuous treadmill surface for both walking and running. Walking data were obtained by having the subject walk with each foot striking one of the side-by-side treadmill force platform units. Running data were obtained using the larger unit. Raw treadmill data were preprocessed using algorithms developed in house and implemented in LabView before being down sampled and combined with motion capture data (24,25). The preprocessing algorithms detected gait cycle events, initial contacts and toe-offs, for each foot during both walking and running.

Knee joint torque data in the sagittal, coronal, and transverse planes were calculated through a full inverse dynamic model implemented using Vicon Plug-in Gait (version 2.0). Joint torques were normalized by body weight and barefoot height and reported in newton-meters per kilogram-meters.

From each of two 15-s running trials, nine consecutive cycles of gait were averaged for a total of 18 cycles for both conditions. During walking, four consecutive cycles from each of two 15-s trials for both conditions were averaged. Average curves of knee joint torques were graphed over the gait cycle (0–100%). Maximum and minimum torque values at characteristic peaks during stance were obtained from each subject’s average curves. A secondary examination was performed of knee torque values in the sagittal and transverse planes, and of foot progression angle in the transverse plane. Transverse plane foot progression angle was used to quantify the amount of toe-out during walking and running. Comparisons between conditions were made using paired-samples t-tests. Statistical significance for the primary analysis of knee varus torque was defined at P < 0.05. Applying a Bonferroni adjustment to all secondary comparisons for the use of multiple t-tests six variables during walking and four during running, statistical significance was defined at P < 0.005 (0.05/10).

RESULTS
Graphs of the group mean coronal plane knee torque in control shoes with and without arch support cushions are illustrated in Figure 2A (walking) and 2B (running) over an averaged gait cycle. Peak knee varus torque was significantly greater for arch support cushions in stance during running and in late stance during walking (Table 2). No significant change was observed in the early stance peak of knee varus torque during walking.

Of all exploratory parameters included in the secondary analysis, only peak internal rotation torque during running

<table>
<thead>
<tr>
<th>TABLE 2. Knee torque parameters.</th>
<th>Walking</th>
<th>Running</th>
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<tr>
<td></td>
<td>Control Shoe</td>
<td>Arch Cushion</td>
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<tr>
<td>Primary parameters</td>
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<tr>
<td>Peak varus torque, early stance</td>
<td>0.31 ± 0.06</td>
<td>0.31 ± 0.07</td>
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<tr>
<td>Peak varus torque, late stance</td>
<td>0.20 ± 0.06</td>
<td>0.21 ± 0.07</td>
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<td>Peak varus torque</td>
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<tr>
<td>Secondary parameters</td>
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<tr>
<td>Peak external rotation torque</td>
<td>0.06 ± 0.03</td>
<td>0.07 ± 0.04</td>
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<tr>
<td>Peak internal rotation torque</td>
<td>0.12 ± 0.03</td>
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Mean (± standard deviation) values of select knee torque parameters normalized to body mass and barefoot height (N m (kg m⁻¹)). All comparisons include 95% confidence intervals (CI). * P < 0.05 significant for primary parameters; applying a Bonferroni adjustment for multiple tests, P < 0.005 significant for secondary parameters.
was significantly greater for arch support cushions (Table 2). In contrast to the changes in knee varus and internal rotation torques induced by arch support cushions, no significant effects on sagittal plane knee torque were observed during either walking or running. Furthermore, there were no significant changes in transverse plane foot progression angle during walking (control: 2.7 ± 3.7°; arch cushions: 1.9 ± 3.5°, P = 0.017) or running (control: 5.4 ± 4.7°; arch cushions: 5.0 ± 4.7°, P = 0.016), indicating no change in toe-out angle between conditions.

DISCUSSION

As hypothesized, the addition of arch support cushions significantly increased knee varus torque during both walking and running. Peak knee varus torque during running was increased by a magnitude of 4% (0.03 ± 0.03 N m kg⁻¹) compared with the control shoe alone. This peak value was increased by 6% (0.01 ± 0.02 N m kg⁻¹) in late stance during walking. The increases found during both walking and running were noticeably consistent between conditions, and indicate that arch support cushions exaggerate compressive loading on the medial tibiofemoral compartment. In prior studies assessing the role of footwear on knee varus torque during walking, we reported 23% and 26% increases in both early and late stance caused by high-heeled shoes (2.8-inch height), and a 14% increase in late stance caused by moderate heels (1.5-inch height). During running, Williams et al. showed that aggressively inverted (15–25°) graphite orthotics significantly increase knee varus torque by as much as 27% (31). The present study addressed a more subtle change in footwear. Consequently, the smaller magnitude increases in knee varus torque observed for arch support cushions were not unanticipated, and are still clinically notable. Given that lateral-wedged orthotics are often prescribed for patients with medial compartment knee OA, we previously showed that a 5° lateral shoe wedge reduces knee varus torque by a significant 5% and 7% in early and late stance, respectively (17). Similarly, Crenshaw et al. found in subjects with no knee OA that a lateral shoe wedge reduces knee varus torque by approximately 7% (4). The amount of change observed in this study is likely to be clinically significant, given that changes of similar magnitude for lateral-wedged orthotics during walking have been associated with clinical improvements (13). Particularly so during running, for which medial compartment loading is of a significantly greater magnitude than that during walking, the 4% increase in knee varus torque is of extreme clinical relevance.

Despite results presented by Baker et al. that show limited to no reduction in knee pain in patients with medial knee OA when treated with lateral wedged orthotics, the clinical implications of the present findings cannot be dismissed (2). Although the wedged orthotics Baker et al. investigated were not associated with a change in symptomatic presentation, the same orthotics did have a significant effect on knee varus torque. Additionally, while research contends the connection between knee varus torque and knee OA, given that varus torque is related to coronal plane knee alignment, it is important to note that recent research has questioned coronal plane static knee alignment as a predictor of incident medial compartmental knee OA (11). However, Hunter et al. support that static alignment be used as a marker of disease progression, and concede that malalignment measures provide no impression of the causative nature of increased knee varus torque measured dynamically during gait on the development of knee OA.

Increasing toe-out angle during walking decreases the moment arm of the coronal plane knee torque, and has been proposed as a possible mechanism of knee pain relief (3). A complete assessment of the effects of footwear on knee varus torque, then, must consider a decreased toe-out angle as a possible mechanism for any observed increase in varus torque. The finding that no change in foot external rotation during either walking or running resulted for arch support cushions eliminates the possibility that the increase in knee varus torque was the consequence of a decrease in toe-out angle.

While the methods used in this study are considered to be the most technologically advanced noninvasive techniques available to assess the biomechanics of walking and running, a limitation of our study, and of noninvasive gait analysis in general, is that we must infer rather than directly measure joint contact forces from measured net joint torques. Biomechanical modeling provides support for this inference, showing that differences in net knee varus torques are the major determinants of differences in medial and lateral tibiofemoral compartment contact forces (27). Furthermore, while providing subjects with standardized footwear allowed us to more purely evaluate the effect of arch support cushions alone, we recognize that in studying the effects of footwear on the mechanics of running, not accounting for functional differences in subject-specific foot mechanics prevents characterizing these findings with regard to particular foot types. Finally, research has shown that the effects of shoe interventions on knee varus torque are enhanced in subjects with greater peak knee varus torque prior to intervention (7). The healthy, physically active population assessed in this study presented with a normal range of knee varus torques, and, as such, the effects of arch support cushions on patients diagnosed with knee OA or on those with high knee varus torques could be much greater.

As a relationship is elucidated between footwear and knee varus torque, future epidemiologic studies which evaluate the effect of footwear on the predisposition for knee OA become ever more important. In addition, with the popularity of running as a convenient method of exercise, and given the high magnitude of repetitive lower-extremity loading involved, discretion must be employed with regard to the prescription of orthotic insoles. Furthermore, it is
important to rationalize the impact of over-the-counter inserts on clinical care. The possibility that even subtle changes in footwear influence factors related to the predisposition for knee OA must instigate further biomechanical study in this area, because footwear represents an easily modifiable factor for this prevalent and disabling disease.

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REFERENCES


