Original Research

The Effect of Running Shoes on Lower Extremity Joint Torques

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Objective: To determine the effect of modern-day running shoes on lower extremity joint torques during running.
Design: Two-condition experimental comparison.
Setting: A 3-dimensional motion analysis laboratory.
Participants: A total of 68 healthy young adult runners (37 women) who typically run in running shoes.
Methods: All subjects ran barefoot and in the same type of stability running footwear at a controlled running speed. Three-dimensional motion capture data were collected in synchrony with ground reaction force data from an instrumented treadmill for each of the 2 conditions.
Main Outcome Measurements: Peak 3-dimensional external joint torques at the hip, knee, and ankle as calculated through a full inverse dynamic model.
Results: Increased joint torques at the hip, knee, and ankle were observed with running shoes compared with running barefoot. Disproportionately large increases were observed in the hip internal rotation torque and in the knee flexion and knee varus torques. An average 54% increase in the hip internal rotation torque, a 36% increase in knee flexion torque, and a 38% increase in knee varus torque were measured when running in running shoes compared with barefoot.
Conclusions: The findings at the knee suggest relatively greater pressures at anatomical sites that are typically more prone to knee osteoarthritis, the medial and patellofemoral compartments. It is important to note the limitations of these findings and of current 3-dimensional gait analysis in general, that only resultant joint torques were assessed. It is unknown to what extent actual joint contact forces could be affected by compliance that a shoe might provide, a potentially valuable design characteristic that may offset the observed increases in joint torques.

INTRODUCTION

Knee osteoarthritis (OA), which occurs symptomatically in approximately 6% of adults older than the age of 30 and in 10% of adults older than the age of 55, accounts for more disability in the elderly than any other disease [1,2]. Previous studies on the effect of physical activity on propensity to OA have been conflicting [3-8]. The possibility that the use of different types of footwear during physical activity may 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.

In earlier studies, the authors showed that women's high-heeled shoes increase the external knee varus torque and prolong the external knee flexion torque compared with barefoot walking [9,10], changes that are evident with the addition of even a moderate heel in women's dress shoes [11]. These findings were interpreted to be particularly pertinent given the relatively high incidence of knee OA in women. An increase in the external knee varus torque implies relatively greater compressive force through the medial aspect of the knee, the anatomical site of the knee most prone to degenerative changes, as compared with the lateral aspect [6,12,13].

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Table 1. *Subject parameters*

	n	Height (m)	Mass (kg)	Age (yrs)	Running Speed (m · s^{-1})
Female	37	1.68 ± 0.06	60.0 ± 6.3	31.5 ± 10.3	3.0 ± 0.4
Male	31	1.77 ± 0.07	71.9 ± 7.3	36.8 ± 11.9	3.3 ± 0.4
Combined	68	1.72 ± 0.08	65.6 ± 9.0	34.0 ± 11.3	3.2 ± 0.4

Although the effect was less substantial than those observed in women's dress shoes, men's dress shoes and sneakers also were found to increase the knee varus torque [14]. Similarly, the current authors [15] and others [16] have found that the addition of material under the medial aspect of the foot, such as excessive arch supports or medial wedges, also serves to elicit a medial force bias in gait, thereby increasing the external knee varus torque. Moreover, this increase is further elevated when examining the effect of arch supports during running [15]. Although the authors' previous studies have evaluated the effect of shoes primarily during walking, they sought in the present study to evaluate the effect that typical modern-day running shoes have on these same knee joint torques. Such information could be useful in guiding the prescription and possibly the design of running shoes.

Elevated repetitive loading, such as that experienced during running, is believed to be an important etiological factor in the development of OA [17]. In fact, in animal models, the mechanical stress experienced during strenuous running has been shown to induce degeneration of articular cartilage at the knee [7]. It has long been assumed that running shoes minimize these mechanical stresses [18-20]. Although the immediate health benefits of running are substantial and well recognized, there is no clinical evidence to support that the design of modern running footwear is most favorable to promote long-term health in runners [21].

Indeed, although it is recognized that the typical cushioned running shoes may alleviate actual joint contact forces, the authors hypothesize that certain attributes of running shoe design increase the relative distribution of these forces. Specifically, typical running shoes are designed with characteristics analogous to those previously identified to increase knee joint torques in walking [9-11,14-16]. Current cushioning technologies in running shoes serve to elevate the heel compared with the forefoot. Further, motion control and stability technologies inherent in running shoe design essentially provide additional material under the medial aspect of the foot, via medial posting and arch supports. The authors hypothesize that the contribution of these design characteristics in running shoes would serve to increase both the external knee flexion torque and the external knee varus torque. This study examined the effect of standard athletic footwear on lower extremity joint torques during running.

METHODS

Sixty-eight healthy runners (36 women) were recruited from the local population. Subjects had no history of musculoskeletal pathology and were without musculoskeletal injury at the time of testing. Each subject described his or her running as recreational and ran a minimum of 15 miles each week. The experimental protocol was approved by the Institutional Review Board for Health Science Research, and written informed consent was obtained from each subject. There were no gender effects observed in any of the examined measures, and thus the analysis comprised all subject data (Table 1).

Standardized running footwear was provided to each subject. Although more cushioned or motion control footwear may have been more optimally suited for the foot characteristics of a portion of the study population [22], the control shoe used in this study was the Brooks Adrenaline (Brooks, Bothell, WA), selected for its neutral classification and design characteristics typical of most running footwear. Specifications for the control shoe are compiled in Table 2. For both the shod condition and for barefoot running, subjects were asked to run on an AMTI compound instrumented treadmill (AMTI, Watertown, MA) at their self-selected comfortable shod running speed and complete a 3- to 5-min warm-up period. The acclimation period duration was selected based on previous work concluding that such a duration was appropriate to produce stable estimates of kinetic parameter mean values during treadmill running [23].

Sixteen retro-reflective markers were placed over the following anatomical landmarks of the pelvis and lower extremity: bilateral anterior and posterior superior iliac spines, lateral mid-thighs, lateral femoral condyles, lateral mid-shanks, lateral malleoli, second metatarsal heads, and heels [24]. Excluding the markers placed over the heels and second metatarsal heads, all marker placements were unchanged between conditions. Markers placed on the heels and second metatarsal heads during the shod condition were placed on the shoe over the anatomical landmarks located via palpation. The 3-dimensional positions of each marker were captured at 250 Hz by the use of a 10-camera Vicon 624 motion analysis system (Vicon Peak, Lake Forest, CA).

Ground reaction force (GRF) data were obtained at 1000 Hz from the instrumented treadmill in synchrony with the motion capture data. The characteristics of the instrumented treadmill have been reported in detail elsewhere [23,25]. In brief, it consists of 2 side-by-side forceplate units (330 mm ×

Table 2. *Footwear specifications of the control shoe*

Last shape	Semi-curved
Posting	Dual density
Lasting	Stroebel board
Shore (midsole/posting)	59/(63/70)
Cushioning	Hydroflow
Midsole height, mm (rearfoot/forefoot)	24/12

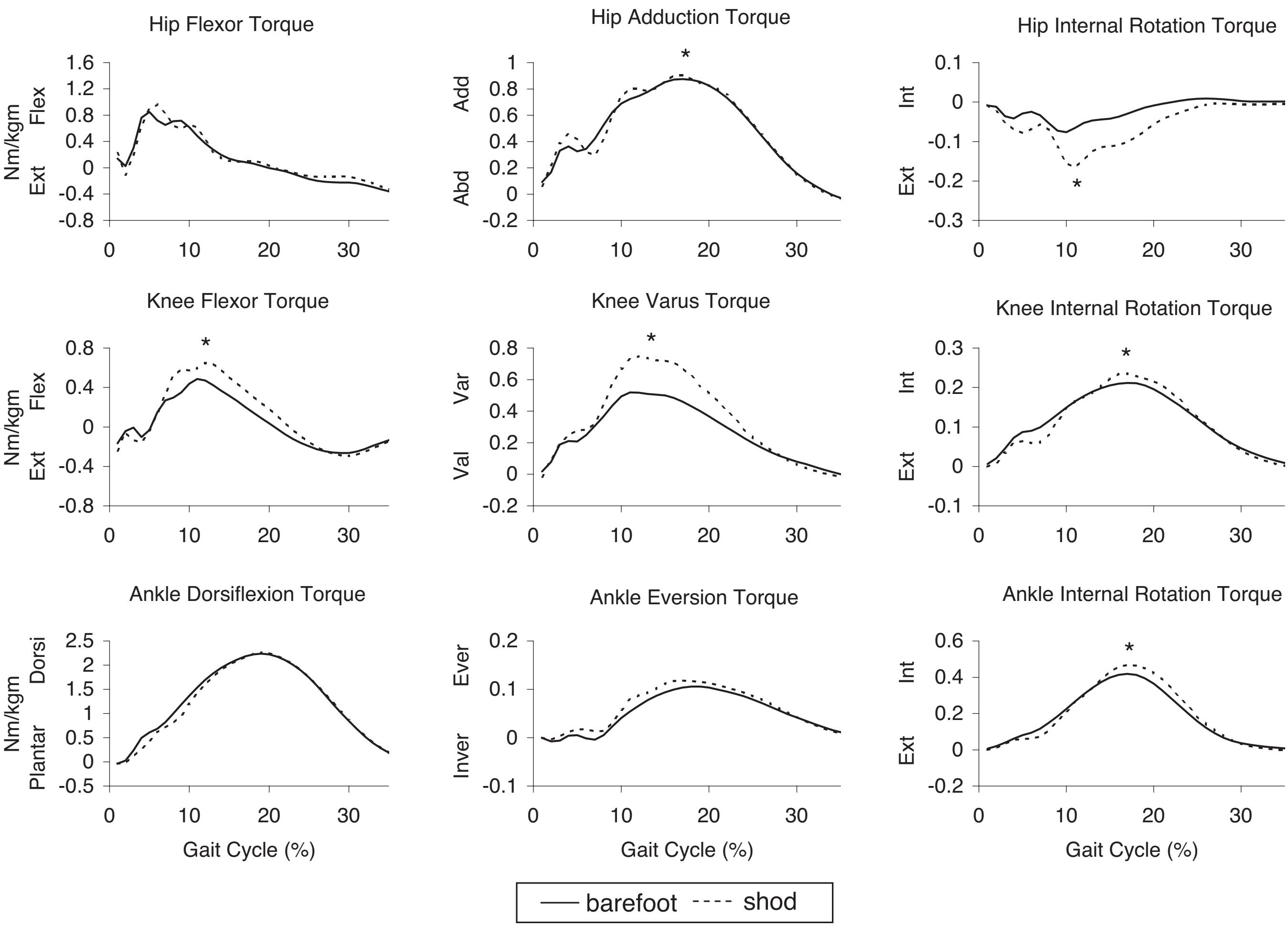

Figure 1. Group-mean sagittal, coronal, and transverse plane joint torques at the hip, knee, and ankle plotted during stance for barefoot (solid line) and shod (dotted line) running. *Indicates significant differences between barefoot and shod conditions, $P < .0035$.

1395 mm) positioned behind a larger unit (662 mm × 2750 mm) providing a continuous treadmill surface. The similarity between overground gait and both walking and running on the instrumented treadmill used in this study has been established previously [23,25].

Running data were obtained by use of the larger unit because the existence of a flight phase allowed consecutive strides to be collected from a single treadmill forceplate. The measured vertical natural frequency of this larger treadmill unit was 219 Hz. For each condition, two 15-second trials of synchronized motion capture and treadmill GRF data were collected. The first of these trials was used in the analysis in the absence of significant marker dropout, in which case the second trial was used. Treadmill force plate data were low-pass filtered at 30 Hz by the use of a second-order Butterworth filter before being down sampled and combined with the motion capture data. Temporospatial parameters were calculated from characteristic events, heel-strikes, and toe-offs, identified analytically for each trial by the use of a 60-N threshold of the vertical GRF. This threshold was selected based on the authors' previous work identifying gait cycle events in the presence of noise levels greater than those apparent in GRF signals obtained from standard static force-plates [23].

Joint torques in the sagittal, coronal, and transverse planes during the stance phase of gait were calculated bilaterally over the course of 10 consecutive strides for each running condition through a full inverse-dynamic model implemented by the use of Vicon Plug-In Gait. Joint torques at the hip, knee, and ankle were resolved in the reference system of the proximal segment, were normalized by body mass and barefoot height, and were reported as external torques. Average curves of the 9 examined joint torques and the 3 components of the GRF were normalized to the gait cycle (0%-100%) and graphed over the stance phase of gait. Subject average maxima and minima during stance were extracted from the each subject's average curve of 10 consecutive gait cycles.

The average overall cycles included in the analysis was reported for each condition. Differences in peak joint torques, peak GRFs, and stride length were assessed by the use of paired-samples *t* tests, and 95% confidence intervals

Table 3. *Kinetic parameters*

	Barefoot	Shod	95% Confidence Interval	*P* Value
Hip flexion torque (Nm · $kg^{-1}m^{-1}$)	1.18 ± 0.36	1.20 ± 0.41	±0.06	.422
Hip adduction torque (Nm · $kg^{-1}m^{-1}$)	0.98 ± 0.16	1.03 ± 0.17	±0.02	<.001*
Hip external rotation torque (Nm · $kg^{-1}m^{-1}$)	0.13 ± 0.04	0.20 ± 0.05	±0.01	<.001*
Knee flexion torque (Nm · $kg^{-1}m^{-1}$)	0.55 ± 0.22	0.75 ± 0.21	±0.04	<.001*
Knee varus torque (Nm · $kg^{-1}m^{-1}$)	0.60 ± 0.15	0.83 ± 0.16	±0.02	<.001*
Knee internal rotation torque (Nm · $kg^{-1}m^{-1}$)	0.24 ± 0.06	0.27 ± 0.08	±0.01	<.001*
Ankle dorsiflexion torque (Nm · $kg^{-1}m^{-1}$)	2.32 ± 0.42	2.34 ± 0.37	±0.04	.485
Ankle eversion torque (Nm · $kg^{-1}m^{-1}$)	0.18 ± 0.09	0.19 ± 0.07	±0.02	.544
Ankle internal rotation torque (Nm · $kg^{-1}m^{-1}$)	0.45 ± 0.12	0.51 ± 0.13	±0.02	<.001*
AP GRF min (%BW)	29.1 ± 4.8	26.5 ± 4.1	±0.46	<.001*
AP GRF max (%BW)	32.8 ± 5.3	33.2 ± 5.6	±0.88	.364
ML GRF min (%BW)	9.9 ± 3.6	10.9 ± 3.2	±0.66	.002*
Vertical GRF max (%BW)	229.8 ± 27.7	238.9 ± 25.8	±2.90	<.001*

Mean (± SD) values of select lower extremity kinetic parameters. All comparisons include 95% confidence intervals. AP = anterior-posterior, ML = medial-lateral. *Indicates significant difference between barefoot and shod conditions, $P < .0035$.

were calculated and reported for all comparisons. Applying a Bonferroni adjustment for the use of multiple *t* tests, statistical significance was defined as $P < .0035$ (0.05/14). For each peak joint torque found to significantly differ between barefoot and shod running, Pearson correlation coefficients were calculated to establish the amount of variance in the change between conditions that could be attributed to changes observed in temporospatial parameters.

RESULTS

Joint torques at the hip, knee, and ankle in the sagittal, coronal, and transverse planes are illustrated in Figure 1. Shod running was associated with increased peak torques at each of the 3 lower extremity joints compared with barefoot running (Table 3). The most prominent increases were observed at the hip and knee. Specifically, disproportionately large increases were observed in the hip internal rotation torque and in the knee flexion and knee varus torques. Although far less substantial, running shod also was found to significantly increase the adduction torque at the hip, and the internal rotation torques at both the knee and ankle as compared with barefoot running. Further, differences between barefoot and shod running were observed in each of the 3 components of the GRF (Table 3). Shod running was associated with a reduction in the propulsive peak of the anterior–posterior GRF and increases in the peak medial–lateral and peak vertical ground reaction forces.

Although identical running speeds were maintained between conditions by study design, subjects adopted a significantly longer stride length shod (2.29 ± 0.29 m) than was observed barefoot (2.15 ± 0.32 m) ($P < .001$). However, this increase in stride length was found to have only weak correlations with the increased joint torques observed for shod running (Table 4). In fact, of the 3 disproportionately large increases in joint torques, only the increase in the external knee varus torque was found to significantly correlate with the increase in stride length, still only explaining approximately 8% of the variance in the increase while running shod. While there were statistically significant correlations in each of the GRF components, only the vertical GRF was found to have a strong correlation.

DISCUSSION

These findings confirm that one effect of the typical construction of modern-day running shoes is to increase joint torques at each of the 3 lower extremity joints. These increases are likely caused in large part by an elevated heel and increased material under the medial aspect of the foot because the current authors [9,11,15] and others [16] have previously shown that these 2 footwear characteristics each independently contribute to increasing knee joint torques. The observed 36% increase in the knee flexion torque with running shoes potentially increases the work of the quadriceps muscle, increases strain through the patella tendon, and increases pressure across the patellofemoral joint [26]. Furthermore, a 38% increase in the knee varus torque implies relatively greater compressive loading on the medial tibiofemoral compartment, an anatomical site prone to degenerative joint changes, as compared with the lateral compartment [27-30]. Finally, the disproportionately large 54% increase in the hip internal rotation torque may have particularly high clinical relevance, given previous findings that indicate that competitive running may increase the risk of OA of the hip joint [8]. The running shoe used in this study is manufactured with

Table 4. *Pearson correlation coefficients between change in stride length (ΔSL) and changes in kinematic parameters*

ΔSL vs Parameter	R	R^2	*P* Value
Δhip adduction torque	0.21	0.04	.09
Δhip internal rotation torque	0.21	0.04	.09
Δknee flexion torque	0.17	0.03	.17
Δknee varus torque	0.29	0.08	.02*
Δknee internal rotation torque	0.32	0.10	.01*
Δankle internal rotation torque	0.26	0.07	.04*
ΔAP GRF min	0.35	0.13	<.01*
ΔML GRF min	0.28	0.08	.02*
Δvertical GRF max	0.59	0.35	<.01*

*Indicates a correlation found statistically significant, $P < .05$.

characteristics typical of most running footwear, suggesting that these observed kinetic changes result from the commonly accepted design of modern-day running footwear and are likely not specific to any one manufacturer.

Remarkably, the effect of running shoes on knee joint torques during running (36%-38% increase) that the authors observed here is even greater than the effect that was reported earlier of high-heeled shoes during walking (20%-26% increase) [9-11]. Considering that lower extremity joint loading is of a significantly greater magnitude during running than is experienced during walking [31], the current findings indeed represent substantial biomechanical changes. It is likely that the shoe characteristics previously found to increase knee joint torques in walking are similarly responsible for much of the increases in knee torques presently observed in running. These design characteristics were found to significantly alter the GRF components and subsequently increase the resulting joint torques. However, given the substantial increases, there may be other factors as well. In this study, despite controlling for running speed, the authors did note a 6% increase in stride length, which is consistent with other studies investigating the effects of footwear [32,33]. However, this slight increase in stride length, likely deriving from shoe characteristics that promote foot comfort, was found to account for only a very small portion of the increase in joint torques while running shod.

The design of current running shoes, with various heel-cushioning strategies and technologies to increase medial support to control foot pronation, has become widely accepted as the industry standard. However there is no clinical evidence to support that this design is optimal to promote the long-term health of runners [21]. In fact, the rate of running-related injury in distance runners has not changed dramatically despite advances in footwear design technologies [31]. Footwear theoretically could provide some level of beneficial compliance similar to that achieved with altering running surface [34], potentially reducing joint contact forces. However, it is unlikely that midsole deformations evident in current running shoes have an appreciable effect on leg mechanics at midstance [35]. Heel cushioning, primarily in place to counter impact forces in running, has limited to no effect at midstance, the time occurrence of peak GRFs, peak joint torques, and presumably peak joint contact forces [31]. Medial posting and arch supports on the other hand may inhibit the natural, potentially beneficial compliance of the foot in transitioning from a supinated to a pronated position near midstance back to a supinated position near toe-off [31]. In contrast, recent research has revealed that positive clinical outcomes accompanied the prescription of custom foot orthoses designed with medial posting in women experiencing running-related overuse injury at the knee [36]. These findings emphasize that although the present study observed dramatically increased lower extremity joint torques with the use of typical running footwear, the individual needs of a runner should ultimately dictate footwear prescription.

Although the methods used in this study are considered to be the most technologically advanced noninvasive techniques available to assess the biomechanics of running, a major limitation of this study, and of noninvasive gait analysis in general, is that the calculated joint torques provide only an estimate of the net difference between the forces on either side of a joint rather than an estimate of the actual joint contact forces. Clearly there is a role for the development of gait analysis technologies that would allow for the evaluation of these contact forces. It is also conceivable that the study subjects adopted a different contact style to minimize a potential increase in impact loading associated with barefoot running. However, every effort was made to ensure subject familiarity with running on the instrumented treadmill and each subject reported feeling comfortable with the barefoot condition. Consequently, the reported increases in lower extremity joint torques are indeed genuine effects of the shod condition. Finally, the foot mechanics of the subject population were not used to identify the specific type of footwear that would be most appropriate for each individual. However, the control footwear used in this study was considered representative, with design characteristics typical of most running footwear. The authors have reported [22] on the alignment and barefoot arch profiles of the subject population in detail elsewhere.

Previous studies on the effect of strenuous physical activity, such as running, on propensity to OA have been conflicting [3-8]. Although increased repetitive loading has been shown to be a critical factor for the degeneration of articular cartilage at the knee, the forces experienced by distance runners have not been consistently found to increase the risk of onset of knee OA. Further, there has yet to have been performed an epidemiological study on the development and/or progression of knee OA that controls for footwear. However, the substantial magnitude of the changes presently observed does expose a potential concern regarding the development of joint injury in runners. The use of athletic footwear in running as a means to protect the foot from acute injury and the potentially debilitating effect of switching to barefoot running on foot health excludes such an alternative. The development of new footwear designs that encourage or mimic the natural compliance that normal foot function provides while minimizing knee and hip joint torques is warranted. Reducing joint torques with footwear completely to that of barefoot running, while providing meaningful footwear functions, especially compliance, should be the goal of new footwear designs.

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