Biomechanics Analysis of Barefoot vs. Shod Running

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ABSTRACT
Each year thousands of runners are injured, many incidents being caused by improper form or attire. Prior studies have shown barefoot running to reduce impact loading, contact time, and stride length by encouraging a forefoot striking (FFS) pattern. The aim of this study is to further analyze and compare mechanical characteristics and effects between barefoot and shod running. Data collected from N=11 participants with Qualisys 3D motion tracking system was used to assess applied foot force, inversion and eversion angles, foot plantarflexion, location of the center of pressure of the foot, ankle moment, in addition to moments, forces, and flexion angles of the knee. When barefoot, statistically significant reductions in several parameters, including impact force and peak knee moment were observed which may reduce the overall risk of injury. No significant increase in eversion was observed. By performing this biomechanical analysis on barefoot running, researchers conclude that barefoot style running results in a safe gait pattern. Future impacts of this research may directly affect current training regimens. Increasing awareness of barefoot running may inspire others to verify this study and determine to what degree running related injuries are preventable.

Key Words: Biomechanics, running, barefoot

INTRODUCTION
Over the years, many studies have been performed on the biomechanics of running, from evaluating different shoe types, to complete kinematic studies of running. More recently, running barefoot or in minimalistic shoes has gained attention for purportedly being more beneficial than traditional shoes. With growing popularity, barefoot running has opened new avenues in research to assess the effectiveness of running with different types, or lack, of footwear. Much research has been focused on various biomechanical aspects of running, some of which have provided the scientific community with ample data analyses and are reviewed within this section.

With a focus on the trajectory of the center of pressure during barefoot running, a recent study by De Cock et al. sought to analyze and interpret the center of pressure path during barefoot running as well as determine if foot structure played a role in the center of pressure. It was discovered that there was an increase in the center of pressure length during initial contact and flat foot phases of gait while barefoot [1]. The researchers concluded that the height of the arch contributes to a shift in the location of the center of pressure. While, a lower arch corresponded to a lateral shift in the center of pressure, indicating a slower loading of the metatarsals. This can be analyzed further to determine whether a specific center of pressure pattern correlates with injury.

Similarly, a study conducted by Thompson et al. focused on the differences between impact accelerations of barefoot and shod running. Within this study, one hypothesis was to determine if there was a difference in peak impact magnitudes between the two conditions. This was a heterogeneous study consisting of ten subjects, five males, and five females, all of which were healthy active runners. Participants were asked to run over the top of a runway containing a force plate, in three separate conditions: barefoot, shod, and barefoot heel strike. Measurements were taken using a high speed 3-D analysis camera utilizing 16 markers placed on the participant. Researchers found there was indeed a difference between barefoot and shod running, ultimately concluding that there was a reduction in ground reaction force (GRF) in natural, forefoot strike barefoot running conditions [2].

Changes in foot strike pattern, along with inversion-eversion of the foot, when progressively transitioning to barefoot running over a 12-week period were assessed in a study by Latorre-Román et al. demonstrating inversion and eversion of the foot were both limited significantly when measured immediately as the foot made contact with the ground. After receiving training, at low speed, there was a 33.4% reduction in the number of participants landing with an inverted foot, with these participants then exhibiting a centered strike upon landing. Additionally, changes in external foot rotation were also observed [3]. Further, an alternative study addressed the inversion and eversion of the foot during three phases: foot flat, mid-stance, and toe-off. Here, negative angles correspond to inversion while positives indicate eversion. In the frontal plane, maxima of -18.5° ± 5.42° for inversion and 1.2° ± 2.81° for eversion in the right foot were reported. For the left foot, maxima of -16.8° ± 3.62° for inversion and 0.7° ± 1.05° for eversion were reported [4].
Other aspects of the ankle were addressed in a study conducted by Lieberman et al., in which ankle angles at the moment of foot strike for five separate sample groups with different lifelong running habits were measured. The study was heterogeneous and composed of 44 males and 19 females. Members of each group ran between five and seven barefoot and shod trials along a 20-25 meter track. Kinematic markers were placed at the knee, ankle, and first metatarsal head while Fastec Imaging with force plates was used to gather angle data at the moment of foot strike. In all five groups, barefoot conditions resulted in increased plantarflexion at initial contact. Most applicable to the western world, however, the habitually shod group from the USA expressed 9.5° less plantarflexion at the moment of initial contact while running with shoes when compared to barefoot. These results imply a clear trend between barefoot running and plantarflexion similar to the results observed in other studies [5].

Further studying the ankle in 2014, Sinclair tested thirty recreational male runners, having a mean age of 26.21 years. Sinclair found barefoot running to result in significant increases regarding: peak ankle plantarflexion moment, Achilles tendon force, and Achilles tendon force load rate. Drawn from previous work, differences between barefoot or barefoot inspired footwear and conventional running shoe types were considered for the following: knee extensor moment, patellofemoral contact force, patellofemoral pressure, patellofemoral contact force load rate. With each of these, a significant increase was found in the conventional shod state, compared to the minimalistic shod or barefoot state. [6].

Just as, or perhaps more important than the ankle, the knee is an area of extreme interest to runners. German researchers at the University of Cologne conducted a study to determine the effects of footwear on knee and ankle joints while running, including fourteen healthy male endurance runners. All subjects had been running 3 to 4 sessions per week for the past five years and had no neuromuscular or musculoskeletal issues at the time of the study. To assess the effects of footwear on the joints, five different shoe types, along with barefoot, trials were completed. Data was collected with an infrared camera system and retroreflective markers with a force plate and a 20-meter long running track that was fitted with grass-like padding for the natural barefoot condition. Parameters of the study included: ankle angle, knee angle, ground reaction force, gear ratio at the ankle and knee, and moment at the ankle and knee. Participants in this study did not show a significant difference in ankle or knee angle at ground strike and toe-off. However, it was shown that barefoot running exhibited significantly less contact time than shod conditions. The greatest statistical difference occurred with the amplitude of vertical ground reaction force, shod having a higher amplitude. The researchers concluded that wearing shoes affects the moment at ankle and knee joints by extending the moment arm of the ground reaction force. Additionally, knee loading was greater in shod running, which can be more painful for those with osteoarthritis or knee injuries [7].

Another study performed in 2012 had participants run barefoot and shod while forces, angles, and moments about the knee were recorded. Researchers found that when running barefoot, the work output on the knee was greatly reduced whereas the work output on the ankle was increased. The primary purpose for this change was attributed to alteration in running style, namely, forefoot striking while barefoot versus rearfoot striking while shod [8].

After reviewing the literature, conducting a study that focuses on multiple aspects of running could prove to be beneficial. Analyzing multiple parameters on one participant group allows for the observation of how these parameters work in relation to one another. Therefore, the main objective of this study is to explore the hypotheses that running barefoot will result in the following: decreased applied foot force, decreased inversion and eversion angles, increased plantarflexion of the foot at initial contact, different center of pressure length during specific instances of gait, differences in peak torque in the ankle, and decreases in impact forces, torques, and differences in flexion angles at the knees.

II. METHODS

Participant Preparation

The present study consisted of N=11 participants, seven males and four females, all respondents to recruitment flyers placed around Florida Gulf Coast University. Participants ranged from the ages of 20 to 27 years old. The mean characteristics of the participants were: age=22.64, SD=1.96 years, body mass=75.24, SD=16.99 kg. They were required to have had running experience of at least once per week and able to sustain a pace of four miles per hour or greater for at least thirty minutes. Participants were free of acute musculoskeletal or neurological disorders within the six months preceding their participation in the study. Participants were not able to use any additional equipment for their shoes, such as gel pads or prescription orthopedic inserts. This study and recruitment flyer was approved by the Institutional Review Board (IRB) at Florida Gulf Coast University.

Data Collection

Participants were asked to meet in the Biomechatronics lab at Florida Gulf Coast University. Prior to beginning data collection, the participant’s feet were screened for any visible infections, blisters, scrapes, and abrasions. Reflective markers were then placed bilaterally on the lower extremities in the following locations: first and fifth metatarsal heads, medial and lateral malleoli of the ankle, medial and lateral femoral epicondyles, the greater trochanters, and the left and right calcaneus. When starting each trial (barefoot or shod) a total of four runs were conducted: two as warm up and two for data collection. These warm up runs allowed the participant to adapt to the running environment thus minimizing any potential
irregularities in their gait pattern.

All data was measured and collected by the Qualisys Track Manager software (Qualisys Medical AB, Goteburg, Sweden) with a capture frequency of 2000 Hz for force data, and 200 Hz for motion capture. During data collection, participants were instructed to run across the Biomechatronics Lab within the range of the Qualisys. Participants were instructed to run either barefoot or shod for their first run and opposite for the second, where this experimental condition sequence was randomly determined per participant. When running, they were asked to run at a pace at which they would use if they were to run a typical mile. For each running cycle, the participant started outside of 108 feet, prior to having one of their last few strides of the data collection area and ran for a maximum distance experimental condition sequence was randomly determined per participant. All participant data was saved anonymously, then exported and further analyzed in MATLAB, a computational software package in which subsequent calculations were performed.

Center of Pressure Location

The length of the center of pressure (COP) of the foot, at a given time during the stance phase, was determined by the following. The foot was first defined as a vector from the heel to the toe. Then an additional vector, COHeel, was defined as the vector between the heel and center of pressure of the force plate, as recorded by Qualisys Track Manager. Once defined, a dot product between the two vectors was performed, then divided by the magnitude of the foot vector. The output of this calculation is the length of the COP of the foot from the heel, valid for any point in time for which the participant was in contact with the force plate. This value was then converted into a percentage of overall foot length by dividing by the foot vector. Calculations were computed using Equation (1).

\[ \text{CoP \%} = \frac{\text{COHeel} \times \text{Foot Vector}}{||\text{Foot Vector}||} = \frac{||\text{COHeel}||}{||\text{Foot Vector}||} \cos(\theta) \]  

Ground Reaction Force (GRF)

Measured GRF was normalized per participant by dividing the GRF by their body mass in kilograms. This was done to allow for cross comparison between participants. Utilizing this normalized GRF, the striking force threshold was defined by identifying the frames at which ground strike and toe-off occurred. Ground strike occurred once the vertical force exceeded 0.1 N/kg and toe-off occurred once the vertical force dropped below 0.1 N/kg, as performed in another study. These frames were later used to identify all points within the stance phase of gait, and measures were reported per percent of stance phase. This was done for both barefoot and shod running conditions. The maximum normalized GRF was calculated and compared between barefoot and shod conditions, as was this GRF during the loading response phase of stance.

Inversion and Eversion

The amount each participant’s foot was inverted or everted during stance phase was observed based on the position of the first and fifth metatarsal head markers. These angles were calculated by first taking the height of the foot relative to the ground in the z-direction. Next, the relative distance h or height between the first and fifth metatarsal head markers (MH1z and MH5z) was computed in Equation (2). Equation (3) was then used to calculate the angles of the foot relative to the ground. A positive (eversion) angle was defined when the height of the fifth metatarsal head was above the first metatarsal head, while a negative (inversion) angle was defined when the first metatarsal head was higher than the fifth metatarsal head.

\[ h = MH5z - MH1z \]  
\[ \theta = \sin^{-1} \left( \frac{h}{(MH5z-MH1z)^2 + (MH5y-MH1y)^2} \right) \]  

Ankle Angle

To calculate the degree of ankle plantarflexion, the physiological segments of the foot and shank were distinguished by marking the left and right medial ankles (LMA and RMA), knees, and first metatarsal heads. The RMA and right medial knee (RMK) markers signified the endpoints of the shank vector while the RMA and right big toe (MH1 or RBT) markers defined the endpoints of the foot vector. The aforementioned vectors were input into Equation (4) to yield the ankle angle.

\[ \theta = \cos^{-1} \left[ \frac{\text{Foot Length} \times \text{Shank Length}}{||\text{Foot Length}|| \times ||\text{Shank Length}||} \right] \]  

After performing a dot product between the two segments, dividing by the product of their magnitudes, and taking the inverse cosine, the ankle angle was obtained. Using the force plate as a threshold trigger in the same manner done for GRF, the ankle angle during initial contact was determined. The entire stance phase ankle angle for each trial was measured and plotted versus percent stance.

Ankle Moment

The moment in the ankle joint was analyzed by defining axes, velocities, and accelerations of the involved segments were computed via Equation (5).

\[ \text{Ankle moment} = I_{\text{foot}} \alpha_{\text{foot}} - (\text{COP} - \text{COM}_{\text{foot}}) \times \text{GRF} - (\text{Ankle} - \text{COM}_{\text{foot}}) \times \text{Force}_{\text{ankle}} \]  

Where the moment of inertia, \( I_{\text{foot}} \), about the center of mass, COM, of the foot was determined with anthropometric data in conjunction with the length of the foot as a vector, as measured by subtracting the toe position from heel position. The dimensionality of the moment was simplified by simply taking the magnitude of the moments about all axes.
Knee Flexion Angle

Knee flexion angle calculations were computed using Equation (6) along with data acquired from the Qualisys System. Angle theta, $\theta$, was calculated with a specific focus on the angle between the posterior portion of the shank and thigh.

$$\theta = 180 - \cos^{-1} \left( \frac{[TSL + SSL]}{|TSL||SSL|} \right)$$  \hspace{1cm} (6)

Where thigh segment, TSL and $|TSL|$, was defined from the marker at the Greater Trochanter to the lateral femoral epicondyles and shank segment, SSL and $|SSL|$, was defined from the markers at the lateral femoral epicondyles to the lateral malleolus, all in the x- and y- plane. These calculations were only applied to the leg that each participant struck the force plate with. All knee angle data was analyzed for the stance phase including the parameters heel strike, toe-off, and peak stance flexion.

Knee Moment

Utilizing markers placed on the ankle, knee and greater trochanter, a local axis for the shank was created in order to acquire meaningful vectors, rather than using the global (lab) positioning frame. The GRF, in conjunction with shank velocities as well as accelerations and anthropometric data, allowed for the determination of moments generated in the knee [9]. This methodology allows for all calculations to be valid for all subjects during stance phase. Computations were conducted using Equation (7).

$$K_{sh} = I_{sh} * a_{sh} - (COP - COM_{sh}) \times (GRF) - (Knee - COM_{sh}) \times Force_{knee}$$  \hspace{1cm} (7)

Statistical Analysis

Statistical computations were performed in MATLAB after arranging data into columns specific to the barefoot or shod condition, including sample means, sample standard deviations, and paired Student t-tests. Although tests for whether the data fit a normal distribution were not performed, means and standard deviations for each outcome are contained in Table 1 of the results section. Depending on the comparison being made per each hypothesis, the dependent t-tests were either single or two-tailed. An alpha level of 0.05 was desired, however, as twelve separate hypotheses were being tested, a Bonferroni correction was implemented in which the significance level is divided by the number of hypotheses tested in an attempt to minimize Type I error. This brought the cutoff alpha value determining the significance of each separate test to $\alpha = 0.0042$.

III. RESULTS

Center of Pressure Location

Figures 1 and 2 depict COP location per percent stance as the percent of foot length, for all participant runs, in barefoot and shod conditions, respectively. These graphs provide a clear visual distinction of the COP location at various times throughout stance phase, as measured from the heel. Considering both barefoot and shod running conditions, not only was there a significant difference in the center of pressure at initial contact ($P<0.001$), but there was considerably more variability at this point of the gait cycle for the barefoot condition. At midstance, typically taken to occur at thirty percent of the gait cycle, there was also a significant difference between the two conditions ($P<0.001$). The shod conditions exhibit a more consistent pattern between subjects in a percentage of foot length to the center of pressure, seen in Figure 2.

Ground Reaction Force

The body weight normalized GRF from the barefoot and shod conditions are visualized in Figures 3 and 4, respectively. These graphs depict the magnitude of the GRF while over the striking threshold for all participant runs per percent of stance.
No significant difference was determined between the barefoot and shod GRF peak magnitude (Table 1). However, by inspection one can observe the degree to which the experimental conditions differ, specifically regarding an area of additional applied force for the shod condition, near the ten percent stance mark. This additional peak, representative of the loading response phase, was determined to have significantly \((P<0.001)\) lower GRF magnitude for the barefoot condition (Table 1).

\textit{Inversion and Eversion}

The values of the maximum inversion and eversion angles for both barefoot and shod conditions were averaged for all participants. Positive angles represent eversion of the foot while negatives represent inversion. These angles are depicted during stance phase in Figures 5 and 6, where maximum inversion occurs on the left, and maximum eversion occurs toward the right. No significant difference between conditions was found for maximum eversion of the foot. However, barefoot running was determined to result in significantly \((P<0.003)\) lower maximum inversion of the foot (Table 1).

\textit{Ankle Angle}

Ankle plantarflexion angle at initial contact, while barefoot and shod, can be viewed on the leftmost portions of Figures 7 and 8, respectively. Running condition had a significant effect \((P<0.001)\) on ankle plantarflexion at initial contact, indicating that this angle was greater while barefoot (Table 1).
Figure 8: Shod Angle Between the Shank and Foot Per Percent Stance.

**Ankle Moment and Torque**
The resultant of the three-dimensional moment vector about the ankle for barefoot and shod conditions is evident in Figures 9 and 10, respectively. These results were normalized to bodyweight for cross comparison by dividing the moments generated in a trial by the mass of the participant measured in that trial. No significant differences were found between barefoot and shod ankle moment magnitudes (Table 1), although it is worth noting that the barefoot condition has a smoother, more linear rising slope than is true for shod. This is visible between the ten to twenty percent stance phase of the graphs.

Figure 9: Barefoot Ankle Moment Resultant Per Percent Stance.

Figure 10: Shod Ankle Moment Resultant Per Percent Stance.

**Knee Flexion Angle**
Barefoot and shod knee angles were represented separately in Figures 11 and 12, to clarify distinctions between conditions. Contrasting the conditions emphasizes that barefoot knee angles transition in a much smoother manner than shod knee angles, especially between 0-20 percent stance. No significant effects were determined for knee flexion angle at peak stance, initial contact, nor toe-off with an alpha value of 0.0042. Peak stance flexion and initial contact did have P-values lower than 0.05, however differences between conditions were negligible.

Figure 11: Barefoot Knee Angles Per Percent Stance.

Figure 12: Shod Knee Angles Per Percent Stance.

**Knee Moment**
Considering the moment of the knee in three dimensions, the moment about the z-axis, or internal rotation, should approach zero. A large moment about the z-axis would twist the knee undesirably, creating a potential to tear ligaments and tendons of the joint.

Therefore, much of the rotation within the knee is around the x and y-axes, seen in Figure 13. The knee moments for participants in the barefoot and shod condition are represented in Figures 14 and 15. Each of these graphs display moments about all three rotational axes, with the x-axis colored blue, y-axis colored red, and the z-axis colored green. When observing the graphs, it can be noted that the flexion moment, or moment about the y-axis, experiences the greatest torque. The second greatest moment is the Varus rotation peaking at approximately 0.75 Nm/kg lower
than the flexion moment. Similarly, for the shod moments in Figure 15, smoother more consistent paths can be seen than that of bare runners in Figure 14. However, comparing the magnitude of the moments about all axes resulted in a significantly ($P<0.001$) lower peak moment for the barefoot condition (Table 1).

IV DISCUSSION
Focusing on several parameters of the feet, ankles, and knees within the stance phase of the gait, deviations between barefoot and shod running were identified. Analysis of the knee included flexion angles and moment about the joint. Several studies have made note of biomechanical differences between barefoot and shod conditions, many of which have the potential to influence the rate of injuries amongst runners [3]. The following section addresses aspects of these selected parameters.

**Center of Pressure Location**
Research performed by De Cock et al., shows that center of pressure as a percentage of foot length varies with the stance of the participant [1]. Data collected in the present study agreed with these findings. The average percent length of center of pressure at the initial contact phase showed significant differences ($P<0.0042$) between barefoot and shod conditions (Table 1). By visual comparison of initial contact in the left portion of Figures 1 and 2, the barefoot center of pressure is located closer to the forefoot while the shod center of pressure is located closer to the heel. This difference is expected, likely resultant of different striking patterns amongst individual participants, in either barefoot or shod condition. During mid-stance, the percent length of center of pressure was also determined to be significantly different ($P<0.001$) between barefoot and shod conditions (Table 1). Once again by visual comparison of Figures 1 and 2, a difference can be seen during mid-stance. Where the barefoot center of pressure remains more toward the forefoot and the shod center of pressure beginning to shift toward the forefoot. This difference is primarily due to the forefoot supporting the majority of the body weight during barefoot running, which is considered proper barefoot running form (forefoot strike), yielding a center of pressure further away from the heel and closer to the forefoot.
According to De Cock, the center of pressure is 60 percent of the foot length at the 40 percent stance phase [1]. In this study, the barefoot percent length is considerably larger, about 80 percent of the foot length. The difference between the literature and the collected data could result from the measurement of the foot length. In the present study, the foot length was measured relative to the first metatarsal head while previous studies have measured the center of pressure location relative to the entire foot length.

**Ground Reaction Force**

It is most often found that individuals tend to use a forefoot strike pattern during barefoot running for shock absorption, which reduces applied force [5]. The peak GRF measured in the present study held no significant difference between conditions (Table 1). On the other hand, Figure 3 displays a primarily smooth curve for barefoot GRF, this lack of impact transient and the slow rise in applied force is indicative of the loading response phase of stance. Figure 4, containing shod GRF data, shows this noticeable impact transient followed by the expected primary peak. The magnitude of the force applied during loading response was significantly ($P<0.001$) less while barefoot (Table 1), confirming a diminished or absent impact transient for the barefoot condition. Further confirmation is achieved when comparing the present study with that by Lieberman et al., which found that maximum normalized GRF in shod conditions were 1.5 to 3 times greater than while barefoot [5].

**Inversion and Eversion**

In both barefoot and shod runners, preparation for ground strike initiates inversion of the foot as it approaches initial contact, depicted on leftmost areas of Figures 5 and 6. During midstance, the foot is nearly fixed to the floor and therefore minimal changes occur in inversion/eversion during this time, resulting in the plateau region seen in Figures 5 and 6. In preparation to propel oneself forward at toe-off, the foot begins to evert. Finally, during the toe-off phase, represented in the rightmost portions of Figures 5 and 6, the upward slope indicates a rapid increase in eversion angle. These phenomena relate to the mechanical linkages in the foot necessary for this type of movement.

Moriguchi et al. calculated angles of $18.5^\circ \pm 5.42^\circ$ for inversion and $12^\circ \pm 2.81^\circ$ for eversion [4]. Keene states that the typical ranges for inversion are $0-35^\circ$ and $0-15^\circ$ for eversion [11]. Presently measured inversion angles for barefoot resulted in $16.81^\circ$ and $22.62^\circ$ for shod, while the eversion angles were $42.13^\circ$ and $37.77^\circ$ for barefoot and shod, respectively. In this context inversion angles are reported in absolute value, though when measured are negative. Data falls within the normal range of motion for inversion, but not eversion of the foot, as compared with Keene [11]. Results of the present study suggest that barefoot running significantly ($P<0.003$) decreases inversion of the foot, where the average decreased by 5.80 for the subjects. No significant differences between barefoot and shod running can be concluded regarding foot eversion.

**Ankle Angle**

After comparing the average ankle angles for barefoot and shod trials, it was noted that there was a significant difference ($P<0.001$) in ankle angle at initial contact. The plantarflexion occurring at initial contact in the barefoot trials was on average $30.3^\circ$ greater than that of shod trials. In two similar research studies, the aforementioned difference was measured to be $9.5^\circ$ [6] and $14.82^\circ$ [12]. Despite the magnitude of the angular difference being larger than similar studies, the correlation between barefoot running and increased plantarflexion still held true. Key features of Figure 7, such as amplitudes, intercepts, and curvatures, closely match parallel research studies, particularly the ankle angle figures from a 2004 study by Biewener et al. [9].

**Ankle Moment and Torque**

Sinclair found there were significant increases in plantarflexion moment in the barefoot conditions, as compared against several types of athletic shoes [6]. The present study finds no significant differences in peak ankle moment between the barefoot and shod conditions (Table 1). This is evidenced in Figures 9 and 10, where differences in ankle moment magnitude are not readily apparent.

**Knee Flexion Angle**

When visually assessing Figures 11 and 12, differences between the smoothness of knee angle transitions are easily discernable. When participants ran barefoot, there was a smooth transition from the large knee extension angles expected at initial contact to the increased knee flexion angles expected during transition to mid-stance. Conversely, during the shod condition, participants experienced a slight plateau and sharp, fast decline right after initial contact (between 0-20 percent region of stance phase). This is due to the difference in striking patterns, whereas shod runners generally follow a heel-strike pattern and barefoot runners experience a forefoot-striking pattern. These sharp and fast angle changes may indicate higher shearing rates in the joint since sharper changes in angle over similar periods of time means more movement within the joint. This difference may also indicate a disturbance in a runner’s natural pattern, resulting in higher chances of injury while shod. With an average heel strike, or initial contact, knee angle of $163.5^\circ$ and standard deviation of 5.6, these measures can be deemed accurate because values agree with the study from University of Cologne finding angles of $160^\circ$ with a standard deviation of 4 [7]. No significant differences were found when assessing knee angle during heel strike, toe-off and peak stance flexion. However, peak stance flexion and toe-off did have $P$-values lower than 0.05, and graphs are visually distinct. Future studies could compare knee angle ranges throughout the entire stance phase, rather than a few chosen intervals.

**Knee Moment**

Stated in an article by Bonacci, moments about the knee shift between the knee and ankle [8]. Barefoot running has
shown that the moments, on average, are primarily about the knee and are approximately 10 percent lower than running shod. The present study finds the peak knee moment to be significantly ($P<0.001$) less when running barefoot. This is evidenced in Figures 14 and 15, where the primary contributions to knee moment magnitude are in blue and red, both having visibly smaller amplitudes for the barefoot condition.

Conclusion
The aims of the present study were to quantify, analyze, and compare biomechanical aspects of the knee, ankle, and foot for both barefoot and shod running, while further determining what differences held significance. Researchers found that running barefoot decreased the following: impact force transient, inversion of the foot, and peak moment experienced in the knee. Center of pressure and ankle plantarflexion angle are greater while barefoot. No significant effect was found for: peak impact force, knee and ankle planatarflexion angle are greater while barefoot. No significant effect was found for: peak impact force, knee flexion angles, nor peak moment experienced in the ankle.

If cyclically loading connective tissues while running poses a risk of injury, it ought to be minimized by discovering optimal equipment and methods runners should use. With a barefoot style, forefoot strike pattern, the fascia of the foot are allowed a more natural response rate, potentially leading to increased foot muscle strength [8]. With a heel strike pattern, commonly associated with shod running, the ground reaction force is applied primarily to the calcaneus, leading to an impact force transient, which is absorbed by the thicker heel cushion seen in traditional running shoes. This enclosed environment may limit proprioception, leading to decreases in muscle tone of the foot [5]. Additionally, this elevated heel necessitates a greater degree of plantarflexion to achieve an FFS pattern [3].

While there is an inherent degree of conflicting evidence within the research community, this study concludes that for athletes and recreational runners alike, particularly those inhibited due to knee pain, barefoot or minimalistic footwear may be beneficial due to reductions in peak torque within the knee. Eversion and internal tibial rotation are associated with risk of injury [3], and the present study found no significant increase in eversion angle. However, the transition to barefoot running should be done gradually and with caution. Individuals with foot and ankle vulnerabilities might be wary due to potentially increased force on the small bones of the foot at initial contact resultant of FFS, although no significant difference was found in ankle torque.

Future research could elucidate whether a particular center of pressure trajectory is preferential for reducing the occurrence of running related injuries. The clinical and biomechanical significance of cadence and associated angular velocities about joints are also unexplored areas of concern. It is yet to be determined if proper form and training regimen may be more influential to the runner rather than a choice of footwear, though shoes having a smaller heel cushion elevation should allow for a more natural forefoot or midfoot striking pattern.

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