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Comparison of EMG Latencies of the Tibialis Anterior and Soleus in Barefoot and Shod Conditions During Walking

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COMPARISON OF EMG LATENCIES OF THE TIBIALIS ANTERIOR AND SOLEUS IN BAREFOOT AND SHOD CONDITIONS DURING WALKING

by

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Approved by:

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Introduction

The debate between barefoot walking/running and shod walking/running has been prevalent since the invention of the modern athletic shoe, but has experienced a recent surge in interest. Individuals on each side of the debate are quick to espouse the benefit of their method of movement over the other group. While the majority of the debate is from a subjective approach, there is research being done on the differences between shod and barefoot gait.

Studies published have dissected numerous aspects of the biomechanics of running/walking shod and barefoot. Previously, barefoot groups were often used as the control for the studies, but current studies are paying more attention to the comparison of shod versus barefoot runners. Different biomechanical aspects of the both of these gaits have been examined in order to determine what differences are evident between barefoot and shod individuals. Understanding the differences between these two treatments can provide individuals with information necessary to choose what method of locomotion is most beneficial.

One observation is that the foot strike pattern changes between groups that were shod in these studies as opposed to groups that ran barefoot regardless what type of research was being done. Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, Davis, et al. (2010) specifically studied the kinetics of foot strike in barefoot and shod runners. This research found that individuals who are shod almost always adopt what is known as a rear foot strike (RFS). A RFS occur when an individual's heel lands first because the modern running shoe has more cushioning as well as an elevated heel (Lieberman et al., 2010). This allows the individual to experience a comfortable landing on their heel, so he/she adopts a foot strike pattern that works with the shoe.

Individuals who were habitually shod were predominantly RFS while shod and barefoot, but adopted a flatter foot placement when barefoot. Subjects who grew up barefoot or switched to barefoot running often adopted a forefoot strike (FFS) regardless of whether they were shod or barefoot (Lieberman et al., 2010). A mid-foot strike occurs when an athlete lands on the distal portion of the middle of the foot. A FFS occurs in a similar way, but the athlete instead lands on the distal portion further up the front of his/her foot. Barefoot runners will adopt a more horizontal placement of the foot than athletes who are shod (De Wit, De Clercq, & Aerts, 2000). This horizontal placement is said to be adopted because the subjects are attempting to minimize the pressure that acts on the heel. They adopt a strike pattern that automatically adjusts for the impact because they do not have a cushioned shoe to help with the impact. The feet and lower legs now become the mechanism with which the body deals with the stresses of running. This strike pattern is consistent with how individuals would have run before the invention of the modern running shoe. Both FFS and MFS patterns are characterized by an individual reaching for the ground with the distal portion of their foot followed by heel contact (Lieberman et al., 2010). The plantarflexed ankle position of a FFS and MFS is opposite of a RFS because they have a more dorsiflexed ankle which would create a heel strike.

A RFS greatly increases the ground reaction forces that a runner will experience when his/her heel hits the ground because he/she is experiencing more abrupt braking upon impact. Barefoot runners have been found to have vertical ground reaction forces (VGRFs) that are less than individuals who are shod (Kerrigan, Franz, Keenan, Dicharry, Della Croce, Wilder, et al., 2009). The FFS has been found to have VGRFs that lack a distinct impact transient (Lieberman et al., 2010). This impact transient usually occurs within 60 ms after the heel hits the ground in a RFS. These data show that an individual with RFS could experience up to three times his/her body weight at heel contact, while someone with FFS would not experience these impact transients thus making his/her landing less abrupt. The RFS group experiences these greater impact and impact transients because their legs are forced to stop when the heels come down. The ankle was also more compliant in the barefoot groups, which also aided in the reduction of impact. Even though a FFS pattern has been found to reduce the GRF during running, there are also more changes that occur before impact is actually made with the ground.

According to De Wit et al. (2000), when running unshod an individual must adopt a FFS as opposed to a RFS, but this change must be made prior to the impact. The foot must prepare for the landing as soon as it finishes with the impact phase. During shod running an individual will adopt a much more dorsiflexed positioning of the foot, which indicates that the body is preparing to strike the ground with the heel. On the other hand, a barefoot runner will adopt a more plantarflexed position of their ankle that makes the foot more level with the ground. This relationship between the foot and the ground is an effort on the part of the body to prevent a runner from striking the ground with his/her heel (Kerrigan et al., 2009). It is no longer comfortable to take longer strides and land heel first, so the foot prepares to absorb the impact that was usually taken on by the shoe. By landing with a FFS or MFS a runner can avoid overloading of the heel by spreading the initial ground contact over the larger plantar area. While load is reduced on the heel, other tissues would have to dissipate the shock.   During the stride cycle barefoot runners have been found to maintain a greater amount of ankle planter flexion (McNair & Marshall, 1994). Individuals who FFS have also been found to strike the ground with greater inversion of the foot (Williams, McClay, & Manal, 2000). This rearfoot supination is another way that the foot is preparing for a FFS. This is not a conscious behavior, but the brain senses that positioning adjustments need to be made in order to prevent a heel impact.

Not only does the foot undergo a positioning change before a forefoot strike, but the leg also undergoes adjustments in order to accommodate this stance. The leg must prepare in the same way as the foot does, because the impact will be different than the one experienced through RFS. It has been shown that there are differences in the gear ratio of the knee when the act of running was divided into five phases. The gear ratio examined in the study was the ratio of the moment arm of the GRF to the moment arm of the counteracting muscle unit. The ankle and knee both change their gearing in the first, third, and fifth stage of stance phase (Braunstein, Arampatzis, Eysel, & Bruggemann, 2010). The changes that occur in these phases have been found to occur because of the length of the moment arm of the GRF. The knee will also experience higher flexion velocity that would work to reduce impact loading by reducing the mass of the effective leg (Wright, Neptune, van Den Bogert, & Nigg, 1998). This higher velocity is a strategy utilized to reduce impact and it originates 20 ms before touchdown (De Wit et al., 2000). Since the knee and ankle joints are changing in the pre touchdown phase, this allows the foot to land in a much flatter placement than when compared with a shod RFS.

There are changes in how the leg prepares to land during barefoot running, so the musculature that are being used also must adapt to the loads. A shod RFS is going to use the cushioning of the shoe to reduce the impact caused by the running motion, but a barefoot FFS/MFS needs to use the leg itself as cushioning since the landing area of the foot is not padded. During a RFS the hip and knee are going to absorb the majority of the impact, which is shown by their increase in peak torques (Kerrigan et al., 2009). In contrast to the RFS, the FFS engages the gastrocnemius, Achilles tendon, and ankle in order to dissipate the shock.

Even though much of the research on barefoot and shod groups have found that when individuals are barefoot they will adopt a FFS or MFS, there is research that shows that barefoot individuals do not always adopt this gait pattern. Hatala, Dingwall, Wunderlich, and Richmond (2013) studied a Daasanach tribe in Kenya and found that foot strike was a product of running speed as opposed to whether the subjects were shod or unshod. When the subjects ran between 2.01-5.00 m/s the majority utilized a RFS, but upon increasing speed beyond 5.00 m/s, a MFS was more frequently used. These results illustrate that the question of foot strike is not as simple as just saying that barefoot individuals automatically RFS or FFS and shod individuals FFS. This study is in contradiction with previous studies and makes a case for speed being a large determining factor for foot strike type.

While the majority of the research that is currently being done focuses on VGRF and kinematics there is research being done exploring the concept of muscle tuning. The forces that occur during the stance phase of running are input signals and then the locomotor system uses this information to adapt muscle activity to meet the need of the individual (Nigg, 2001). Muscle activity is adjusted to maintain a movement pattern that is best for the individual subject, which could impact fatigue, performance, and work. Impact forces would just be part of the gait cycle and would not play a role in injury of barefoot or shod individuals.

One area of research that has not been examined as thoroughly in regards to barefoot and shod gait is the initial activation of muscles on foot contact. While numerous articles have been published on electromyography (EMG) activity during gait, they often examine different conditions that may cause gait alteration. Differences have been found in diabetic patients and a control group in barefoot and shod walking trials (Sacco, Akashi, & Hennig, 2010). Shod walking was marked by a higher VGRF that could be because of the lack of sensory information that can lead to a more precautious gait in the barefoot group. The healthy participants in the study also experienced a delay in vastus lateralis activation at initial contact time and likely caused by changes in the plantar surface (Nurse, Hulliger, Wakeling, Nigg, & Stefanyshyn, 2005).

Foot arch structure has been another condition examined to determine if it impacts muscle activation in the lower leg. Murley, Menz, and Landorf (2009), examined if muscle activation is altered in flat arched individuals when compare to those with a normal arch. The maximum EMG amplitude was examined and found to be impacted by foot arch height, where the flat arched group experienced a greater EMG amplitude during contact phase for the tibialis anterior during contact and the tibialis posterior during midstance when compared to normal arched subjects. While it is clear that arch height and pathological diagnoses can impact EMG activity, more research needs to be done to determine the relationships that have been uncovered. Valuable information can be gathered from these studies, but more work that examines muscle activity in barefoot and shod conditions must be performed.

The purpose of this study was to examine the differences in peak latencies of the tibialis anterior and the soleus between shod and barefoot individuals when walking at a self selected pace. Kinematics measures were also incorporated to examine the knee angle at heel contact. It is hypothesized that there will be a difference between the shod and unshod conditions in both muscle activation and knee angle. The activation of the tibialis anterior and soleus was expected to occur later in the shod group when compared to the barefoot group.

Methodology

Participants

Fourteen undergraduate university students (10 males, 4 females) (ages 19-36, 22.1 ± 4.4 yr) were recruited for the study. Participants were healthy and free of any neuromuscular dysfunction or orthopedic corrections. All participants filled out an informed consent and a health history questionnaire before participating in the study. The SIUC Human Subjects Committee approved the study prior to data collection.

Experimental set-up

Kinematics data were captured at 100 Hz using a four-camera Qualysis motion capture system and QTM software (Gotenberg, Sweden). These cameras were located around a 7m runway containing an embedded force platform in the middle of the runway. Kinematics residuals during system calibration were all under 0.99 mm, which was the accepted threshold. Three reflective skin markers (14mm diameter) were attached to boney landmarks on the right lateral maleolus, lateral epicondyle of the femur, and over the greater trochanter to capture sagittal plane motion. These markers were attached using double sided tape that firmly adhered to the marker as well as each subject's skin. The boney landmarks were located through palpation and then the markers were attached.

Surface electromyography (EMG) data were collected using a Motion Labs System (Baton Rouge, LA) MA300 system. Data were collected at a rate of 1000 Hz with a common mode rejection ratio of 100 dB at 60Hz, and an input impedance of 10 MOhms. The bandpass width was 20-500 Hz. Silver-silver chloride pregelled electrodes with a bipolar configuration were placed 2cm apart on the tibialis anterior and soleus of the right leg. An electrode pair for the tibialis anterior muscle was placed 3 cm distal to the tibial tuberosity in the middle of the muscle belly. The electrode pair for the soleus muscle was placed 2.2 cm distal to the intersection of the lateral and medial heads of the gastrocnemius muscle. The two electrodes were placed parallel with the muscle fiber of the tibialis anterior the soleus to ensure a clear signal. The ground electrode was placed on the right iliac crest.

A force platform (AMTI, Boston, MA) with a natural frequency of 300 Hz embedded within the runway was used to collect ground reaction forces. Force data were collected at 1000 Hz when the subjects walked across the force platform. All data were synchronized with an external 12-bit analog to digital (A/D) board and stored for future processing.

Subjects were examined in two different walking conditions,: shod and barefoot. For the shod condition, the subjects were instructed to wear any comfortable athletic shoe that would allow a marker to be placed on the ankle.

There was not a specific requirement as to what type of shoe they were to wear. During the unshod condition subjects were completely barefoot.

Testing procedure

Upon arrival at the biomechanics lab, subjects were given an informed consent form and a medical questionnaire. The researcher discussed the purpose of the study and what would be expected of the subject as well as any potential risks from participating in the study. After subjects finished filling out the paperwork, they performed a 5-minute warm-up on the treadmill. When the subjects completed the warm-up, they were prepared for the study by applying kinematics markers and EMG electrodes. The reflective markers were placed on boney landmarks with double-sided tape as detailed previously. After placement of the markers, the subjects were prepped for the placement of the electrodes. Isopropyl alcohol was applied to a cotton ball and the area was rubbed to remove any residual skin and oil that could impact electrical conduction. The electrodes were placed on the tibialis anterior and the soleus as previously discussed. After the electrodes were in place, the electrodes were secured with surgical tape to reduce any movement artifact that could occur during walking.

The wires from the electrode leads were then connected to a collection box that the subject wore around the waist. After the subject was connected, the strength and clarity of the EMG was checked on the computer monitor to ensure proper placement and gain selection. This was accomplished by having the subject dorsiflex to check the tibialis anterior electrode and plantarflex to check the soleus electrode.

A barefoot standing (calibration) trial was recorded in order to determine the distance between the three markers. This calibration was done on the force platform so that the weight of the subject could also be determined. After the initial trial the subjects were then given a chance to familiarize themselves with the runway and force platform. Walking trials began at the end of the runway and subjects had to contact the platform with their right foot. In order for a trial to be considered successful, the entire right foot had to land completely on the platform. When the subjects felt comfortable with the process, five trials were completed in the shod condition at a self-selected walking pace. If the subject showed any signs of braking or speeding up to hit the platform squarely, that specific trial would be redone. Upon completion of the shod trial, the subjects were given a five-minute break to remove their shoes for the barefoot trial. Just as with the first condition, subjects were given a chance to familiarize themselves with walking across the runway and hitting the force platform with their right foot. Since subjects may not have been accustomed to walking in their bare feet, they were given as much time as necessary for them to feel comfortable. Five successful trials were collected and then the subjects were finished with data collection. The trials were counterbalanced in order to eliminate any order-effect that could have occurred.

Data Collection Procedures

While the subjects performed their five trials in each condition, a digital file was being recorded that captured the location of the three markers. The collection of data was counter balanced between the subjects as previously mentioned. The first subject began with the unshod treatment and finished with the shod treatment. The second subject started unshod and then finished in the shod condition. Each collection was recorded for seven seconds to ensure that the entire walk was collected.

Data Analysis Procedures

After data were collected, the files were examined in the QTM software to determine the foot contact and toe off points. These points were determined through examination of the vertical ground reaction force (VGRF) in the recorded file. Foot contact was the first time that the force platform showed a vertical force being exerted, and the toe-off point was when that vertical force was no longer present. These two points were marked in the software and then the files were exported for further analyses.

Electromyography signals were highpass filtered at 100 Hz, rectified, and low pass filtered at 5 Hz with a fourth order Butterworth filter. Force data and kinematics data were low pass filtered at 5 Hz using a fourth order Butterworth filter. After processing, data were analyzed to determine foot contact time by utilizing VGRF. The latency of muscle activation was determined for each trial as well by examining the EMG signals for both the tibialis anterior and the soleus. Latency was the point in the trial where the highest initial peak of the EMG was noted after foot contact (Figure 1, Figure 2).

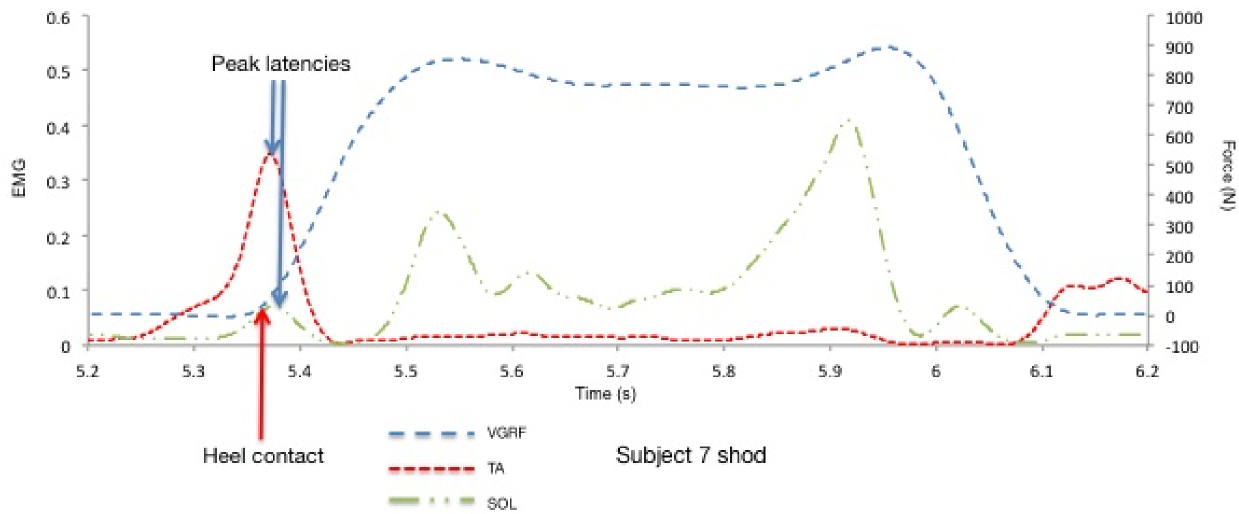

Figure 1. Example of graph used to determine the foot contact time and the peak latencies for the TA and soleus in the shod condition.

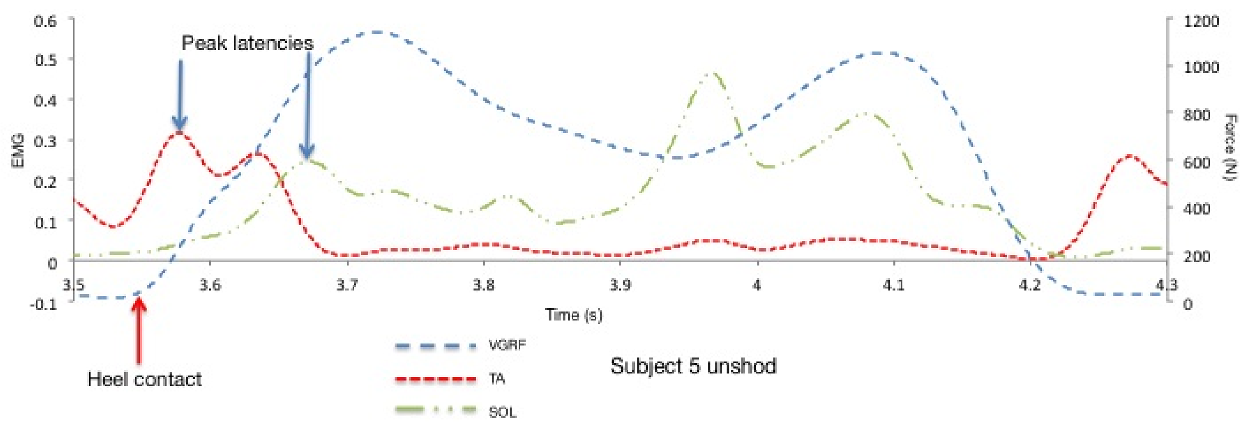

Figure 2. Example of graph used to determine the foot contact time and the peak latencies for the TA and soleus in the unshod condition.

The knee angle at foot contact was determined using the force data from the platform.

Statistical analysis was completed using SPSS. A one-way ANOVA was performed to determine if there were any significant differences in TA and soleus latency between the two conditions. Descriptive statistics were also run to determine if a difference was evident in the mean timing of EMG peak for the TA and soleus in both conditions. A paired t-test was run to determine overall differences between the TA and soleus latency. P-values of less than 0.05 were considered significant.

Results

The first variable that was examined was the difference between the latency of the tibialis anterior (TA) muscle in the shod and barefoot condition. The p-value of this condition was 0.057 with a confidence interval of 95% and an F value of 3.694. While the p-value shows that the values are not statistically significant, the value is very close and exhibits a trend towards significance. The descriptive statistics also show that the mean TA latency for the shod condition was 0.0413 (±0.0272) ms, while the latency for the barefoot condition was 0.0505 (±0.02899) ms.

The second variable that was examined was the latency difference of the soleus between the barefoot and shod group. The p-value of this condition was 0.506 with a 95% confidence interval and an F value of 0.444. There was no significant difference between the conditions for soleus muscle latency. A paired t-test comparing the TA latency and the soleus latency also showed that there was no significant between the two variables (mean difference: -0.04394; 95% CI: -0.05375 to -0.03413).

The mean knee joint angle for the shod condition was 6.913 (±5.0752), and the knee

joint angle for the shod condition was 7.8245 (±5.11522). The p-value of this condition was 0.263, so there was no significant difference in knee angle between treatments.

Discussion

The purpose of this study was to examine the difference between peak latencies of the tibialis anterior and soleus muscle activities during shod and barefoot gait. Kinematics were also collected to determine if there was a difference between the conditions. The hypothesis was that there would be differences between the conditions for both kinetic and kinematics variables. It was found that there were no statistical differences between the two conditions. While statistical significance was set at 0.05, the tibialis anterior p-value was 0.057 showing that a significant trend was present. This finding is of interest when looking at what occurred in the two conditions.

In the shod condition the mean latency for the tibialis anterior was 0.0413 with a standard deviation of 0.0272, but the latency for the barefoot condition was actually 0.0505 with a standard deviation of 0.02899. The shod condition experienced the initial activation peak sooner than the barefoot, which is consistent with other research examining lower limb EMG activity. Scott, Murley, & Wickham, (2012) studied the effects shoe types have on lower limb EMGs, and compared a barefoot condition to a motion control and flexible shoe condition. When subjects were wearing the flexible soled athletic shoe, they experienced a peak amplitude at 5.5% of the gait cycle while the barefoot groups peak amplitude occurred at 6.0%. These researchers believed that this phenomenon occurred because of the mass of the shoe. The extra weight of the shoe would impact how the foot interacts with the ground. That initial peak would indicate the muscle is being activated sooner in the shod group. However there is other research that provides different results.

When individuals are wearing shoes they are altering the information that they are receiving on the plantar surface of the foot, and this alteration of foot feedback has been found to lower muscle activation (Nurse et al., 2005). This effect could be attenuated at foot contact since the sole of the shoe is between the ground and the foot. While not significant the current data show that during the shod trials the shod condition experienced a latency pattern that should be more consistent with a barefoot condition.

This change in activation time could be linked to the primary role of the tibialis anterior which is to provide dorsiflexion of the foot. When the subjects were nearing the force platform, their foot was preparing to touch down by activating the TA and putting the foot in a more dorsiflexed position. Since the time was lessened in the shod condition, they were utilizing the tibialis anterior sooner to pull the front of the foot up for heel contact. The barefoot condition may have experienced a later latency time if they were not utilizing the TA to pull the foot up in as much of a dorsiflexed position. A flatter foot placement would be consistent with previous research showing that barefoot individuals attempt to reduce impact by landing more towards the middle of their foot, so the later latency time could correspond to the difference in foot positioning.

While the TA latency was trending towards significance, the soleus was not significant in the peak latency times between the two conditions. One point of interest when examining the soleus latency is that the shod condition experienced the initial muscle activation peak at a mean time of 0.0929 ms, while the barefoot condition experienced this same peak at 0.0868 ms. The delayed muscle activation of the soleus in the shod condition is opposite of what occurred in the TA. It is possible that the shoes impacted the soleus in a different manner than they had interacted with the TA. While the differences between the soleus and TA can be seen in the descriptive statistics, their lack of significance at the 0.05 level indicates that in the current study the barefoot and shod conditions did not exhibit the same responses as seen in previous studies. The reasoning behind the lack of significance will further discussed in the limitations.

The other metric that was measured was the knee joint angle at foot contact. The shod condition had a mean knee angle of 6.913°(±5.0752), and the barefoot condition had a value of 7.8245° (±5.11522). While these numbers show a difference between the conditions, the p-value of 0.263 is not significant. It is of interest that the knee angle, although not significant, did undergo changes from one condition to the next.

Recommendations

In order to further expand on this research topic, there are limitations that need to be addressed to eliminate the extraneous variables that could impact the results. The first limitation that should be addressed is the footwear that the subjects wear during the trial. The current study did not control for the type of footwear that the subjects wore, so there was a wide variety of footwear. While subjects were instructed to wear athletic shoes, that can mean different things for each individual. Shoes could range from minimalistic shoes to basketball hi-tops, so there needs to be some consistency between what the subjects wear. The simplest way to accomplish this would be to provide all the subjects with the same model of shoe. If they all wore the same footwear, it would be easier to make comparisons between individuals in the shod condition.

Another limitation that would need to be addressed is the speed that the subjects were utilizing for the study. A self-selected walking pace was utilized for the study, so the subjects could have all varied in the pace that they walked. Instead of walking at a self-selected speed for the study, data could be collected on different paces that the subjects are assigned to walk. By using different paces for each subject, each specific pace could be utilized for comparison between subjects. Since each subject would have multiple trails at multiple speeds, it would provide the opportunity to look at the trials of each specific subject as they compare to each other.

Another limitation of the current research study is that current time spent barefoot was not examined. Each of the subjects is coming from a varied background, and the time they spend shod and barefoot will be different. If one is more accustomed to being barefoot, the muscle activation utilized could be different than an individual that is always shod. Since the task of walking barefoot is not novel to them, they will be more comfortable walking which could also impact the results. What could be done in a later study would be to provide a barefoot questionnaire to individuals that will be taking part in the project. The average time that participants spend barefoot could be examined to see if it impacts their EMG when compared to participants who spend the majority of their time shod.

Not only were there limitations in regards to the experimental setup of the barefoot and shod conditions but also the kinematic data that was collected. The only analysis that was done on the kinematics was the knee joint ankle. While this angle was found to be insignificant between trials, it would be valuable to look at what is going on in the rest of the lower leg. It is not possible to examine the angle of contact that the foot is making with the ground without the use of a more comprehensive foot marker set up. By utilizing a more extensive marker set-up on the ankle and foot, one would be able to determine if there is any difference between how the subjects contact the ground. Morio, Lake, Guegun, Rao, and Baly (2009) used a marker set up that utilized 17 distinct markers to determine foot movement. The extensive marker system allowed them to measure plantarflexion/dorsiflexion, abduction/adduction, and eversion/inversion. Sandals were utilized for the study, so footwear would definitely need to be a consideration if this type of protocol is used. Videos of the trials would also be beneficial to determine if there is any difference between the shod and barefoot conditions since it can provide further information on the nature of the foot strike.

Conclusion

Tibialis anterior and soleus peak latency are not affected by whether an individual is barefoot or shod. Knee kinematics were not different between the shod and barefoot condition. While the TA did not show any significance it is notable that there was a trend towards significance. The mean activation time for each condition should be examined further to determine if there is a difference in muscle activation times between conditions. What this research shows it that there is still a copious amount of research that needs to be done revolving around muscle activation and joint angles in shod and barefoot groups. Designing a study that incorporated different speeds of walking and running would allow researchers to examine if these two types of locomotion create the same muscular response. It would also be worthwhile to examine if the VGRF change in relationship to the pattern of muscle activation.

REFERENCES

Braunstein, B., Aramantios, A., Eysel, P., Bruggemann. G-P. 2010. Footwear affects the gearing at the ankle and knee joints during running. Journal of Biomechanics. 43(13), 2120-2125.

De Wit, B., & De Cleercq, D. 2000. Timing of Lower Extremity Motions During Barefoot and Shod Running at Three Velocities. Journal of Applied Biomechanics. 16(2), 169-179.

De Wit, B., De Cleercq, D., Aerts, P. 2000. Biomechanical analysis of the stance phase during barefoot and shod running. Journal of Biomechanics, 33(3), 269-278.

Eslami, M., Begon, M., Farahpour, N., Allard, P., 2007. Forefoot-rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running. Clinical Biomechanics, 22, 74-80.

Hatala, K. G., Dingwall, H. L., Wunderlich, R. E., & Richmond, B. G. (2013). Variation in Foot Strike Patterns during Running among Habitually Barefoot Populations. PLoS ONE, 8(1), e52548. doi:10.1371/journal.pone.0052548

Kerrigan, D. C., Franz, J. R., Keenan, G. S., Dicharry, J., Croce, U. D., Wilder, R. P. 2009. The Effect of Running Shoes on Lower Extremity Joint Torques. PM&R, 1, 1058-1063.

Lieberman, D. A., Venkadesan M., Werbel, W. A., Daoud, A. I., D'Andrea, S., Davis, I. S., et al. 2010. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature, 463, 531-535.

McNair, P. J., & Marshall, R. N. 1994. Kinematic and kinetic parameters associated with running in different shoes. British Journal of Sports Medicine, 28(4), 256-260.

Morio, C., Lake, M. J., Gueguen, N., Rao, G., & Baly, L. (2009). The influence of footwear on foot motion during walking and running. Journal of biomechanics, 42(13), 2081– 2088.

Morley, J. B., Decker, L. M., Dierks, T., Blanke, D., French, F. A., Sterigiou, N. 2010. Effects of Varying Amounts of Pronation on the Mediolateral Ground Reaction Forces During Barefoot Versus Shod Running. Journal of Applied Biomechanics, 2, 205-214.

Murley, G. S., Menz, H. B., & Landorf, K. B. (2009). Foot posture influences the electromyographic activity of selected lower limb muscles during gait. Journal of foot and ankle research, 2, 35.

Nigg, B. M. (2001). The role of impact forces and foot pronation: a new paradigm. Clinical journal of sport medicine: official journal of the Canadian Academy of Sport Medicine, 11(1), 2–9.

Nurse, M. A., Hulliger, M., Wakeling, J. M., Nigg, B. M., & Stefanyshyn, D. J. 2005. Changing the texture of footwear can alter gait patterns. Journal of electromyography and kinesiology: official journal of the International Society of Electrophysiological Kinesiology, 15(5), 496–506.

Sacco, I. C., Akashi, P. M., & Hennig, E. M. 2010. A comparison of lower limb EMG and ground reaction forces between barefoot and shod gait in participants with diabetic neuropathic and healthy controls. BMC Musculoskeletal Disorders, 11(1), 24.

Scott, L. A., Murley, G. S., & Wickham, J. B. (2012). The influence of footwear on the electromyographic activity of selected lower limb muscles during walking. Journal of electromyography and kinesiology: official journal of the International Society of Electrophysiological Kinesiology, 22(6), 1010–1016.

Williams, D. S. III, McClay, I. S., Manal, K. T. 2000. Lower Extremity Mechanics in Runners with a Converted Forefoot Strike Pattern. Journal of Applied Biomechanics. 16(2), 210-218.

Wright, I. C., Neptune, R. R., van Den Bogert, A. J., & Nigg, B. M. (1998). Passive regulation of impact forces in heel-toe running. Clinical biomechanics (Bristol, Avon), 13(7), 521–531.

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