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Gait Alterations and Plantar Pressure in Diabetic Peripheral Neuropathy: A Preliminary Study

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ABSTRACT

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Background: Despite a lack of consensus on its utility, clinicians have traditionally relied on plantar pressure (PP) to predict ulcer risk and prescribe interventions in individuals with diabetic peripheral neuropathy (DPN). Joint kinematics and kinetics have the potential to contribute to DPN assessment and treatment, however previous studies have not accounted for walking speed nor integrated a full-body analysis with a detailed foot model.

Purpose: To assess PP and gait alterations in DPN by controlling walking speed and incorporating a multisegment foot model into a full-body gait analysis. We hypothesize that hip and ankle kinetics will be altered consistent with distal muscle weakness.

Methods: Ten subjects with DPN (height: 178.79 ± 8.55 cm, weight: 108.78 ± 16.67 kg, age: 61.5 ± 13.53 years), and 10 healthy matched controls (height: 180 ± 6.37 cm, weight: 92.87 ±14.5 kg, age: 59.4 ± 7.5 years) participated in this cross-sectional study. Fifty-six reflective markers were attached to each subject according to a full-body model, including a multisegment foot. Subjects walked at a controlled speed (1 m/s) while plantar pressure, kinematic and kinetic data were collected. Functional data analysis was used to compare kinematic and kinetic data between groups, while independent t-tests and a Benjamini-Hochburg procedure was used to compare plantar pressure and joint work metrics.

Results: Individuals with DPN presented with a delayed transition from hip extension to hip flexion moment and a decrease in peak hip flexion moment. There were no major changes found at the knee. There was an increase in peak dorsiflexion angle and delayed power generation in both the ankle and midtarsal joints. DPN subjects also showed a decreased midtarsal positive work. The only significant PP metric found was a decrease in peak PP under the lateral toes.

Conclusion: Findings demonstrated that individuals with DPN use a hip compensation mechanism to overcome distal muscle weakness. Ankle and midfoot alterations are consistent with muscle weakness, requiring proximal compensations. Joint mechanics were more informative than PP measurements and may provide additional insight into DPN assessment and treatment.

Keywords: diabetes, diabetic neuropathy, gait, plantar pressure

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INTRODUCTION

Diabetes mellitus is an increasingly common disease with ultimately devastating consequences. The prevalence of diabetes mellitus has increased worldwide due to a surge in sedentary lifestyles.[1] One of the most feared long-term complications in ambulatory subjects is eventual limb amputation resulting from unchecked foot ulcers. For patients with diabetes the lifetime incidence of developing a foot ulcer may be as high as 25%.[2] A primary cause of foot ulcers is peripheral nerve damage or diabetic peripheral neuropathy (DPN).[2,3,4] DPN causes damage to all 3 types of nerves leading to the distal lower extremity: autonomic, sensory, and motor.[5,6,7] Damage to autonomic and sensory nerves affects sweat production and sensation, increasing the likelihood of plantar skin trauma.[7,8,9] Damage to motor nerves leads to atrophy of intrinsic and extrinsic muscles of the foot, foot deformities, gait changes, and foot loading changes.[7,10,11] These changes can individually and collectively contribute to foot ulceration which may lead to reduced quality of life.

Traditionally, clinicians have used plantar pressure (PP) measurements as a primary means of predicting ulcer risk and prescribing interventions. The connection between PP and ulcers is based primarily on a limited number of cross-sectional studies from the 1970s and 1980s which found increased PP in areas of current or previous ulcers.[12] Several studies have also shown PP differences in DPN compared to healthy controls, predominantly increased PP under the forefoot.[13,14] Interventions based on these studies include offloading devices which aim to reduce the peak PP (PPP) by cushioning the plantar surface and/or redistributing the pressure over a larger surface area.[2] However, studies on interventions based solely off of PP have shown mediocre outcomes[2] and recent research has found only weak correlations between altered PP and ulcer development.[15,16,17] While increased PP may be a contributor to ulcer risk (eg, by

reducing blood flow), it may not be the primary causative factor. In addition, there is still no consensus on specific PP alterations in DPN, optimal PP measurement metrics, or altered PP etiology.[13,14,18]

Additional methods of instrumented gait analysis (beyond PP) have the potential to contribute to DPN assessment and treatment. These have been limited to use with DPN subjects in research settings due to the expensive equipment (eg, motion capture cameras and force plates), spatial requirements, and technical expertise needed. A number of studies have compared DPN gait to healthy matched controls, finding several hip, knee, and ankle kinematic and kinetic alterations. However, 17 of the 20 DPN gait studies relied on self-selected walking speeds,[9,19,20] making it difficult to separate gait alterations due to speed from those due to DPN.[21] Studies using self-selected speeds have been useful in characterizing the slower, more methodical gait used by DPN subjects.[22,23] Out of the many reported gait deviations, it seems only an increase in hip and knee flexion and a decrease in knee extensor moment production are not consistent with decreased walking speed.[21] In addition to unmatched walking speed, previous gait studies have also been limited by modeling considerations. For instance, multisegment foot models have only recently been used to analyze DPN foot movement during gait.[22,20] While these studies have provided valuable information about foot deviations, they have not yet been combined with full-body analyses and have therefore been limited in identifying compensatory mechanisms. Further understanding of DPN gait alterations and compensations may assist clinicians in providing improved assessment and treatment.

Few studies on DPN have incorporated both PP and instrumented full-body gait analysis in the same design. Doing this could potentially help explain changes in PP (eg, a forward shift in weight bearing could cause an increase in PP under the forefoot) and connect DPN gait to

measures being used in the clinic. Therefore, the overall purpose of this study is to assess PP and gait alterations in DPN by controlling walking speed and incorporating a multisegment foot model into a full-body gait analysis. We hypothesize that individuals with DPN will have an increased anterior trunk lean throughout stance. This hypothesis is based on previous research showing general muscle weakness[11,24] and a sedentary lifestyle,[1] as well as increased PP under the forefoot[13] and increased hip flexion during gait.[25,26] We also hypothesize that individuals with DPN will have decreased midfoot and ankle power generation in terminal stance due to atrophy and weakness seen in the smaller distal muscles.[7,10,27] As a result of this, we hypothesize that there will be a compensatory increase in knee and hip power generation.[25]

METHODS

Subjects

A total of 30 participants were recruited from the local community. Due to hardware malfunctions, data from 10 of the subjects were dropped from analysis, leaving 20 subjects for the cross-sectional design. Ten subjects were in the initial stages of DPN (height: 178.79 ± 8.55 cm, weight: 108.78 ± 16.67 kg, age: 61.5 ± 13.53 years). The subjects were screened for and excluded if they had a history of ulcers, amputation, any neurological condition besides DPN, or could not walk unassisted. The other 10 subjects were age, gender and height matched nondiabetic controls (height: 180 ± 6.37 cm, weight: 92.87 ±14.5 kg, age: 59.4 ± 7.5 years). Exclusion criteria included a history of diabetes, any type of peripheral neuropathy, or any lower extremity injury in the past 6 months. All subjects signed an IRB-approved informed consent form before any data was collected.

Protocol

Each subject's height, weight, and date of birth was recorded after which the level of DPN was measured using the Michigan Neuropathy Screening Instrument (MNSI).[28] A total of 56 reflective markers were affixed to each subject with double-sided tape (Figure 1). Markers were placed on the head, acromioclavicular joint line, sternum, the seventh cervical spinous process, medial and lateral aspect of the elbow and wrist, and one marker on the hand. A marker cluster was placed on the posterior pelvis with anterior superior iliac spine and posterior superior iliac spine landmarks identified using a digitizing pointer. Additional clusters were used to track the thigh and shank with individual markers on the medial and lateral aspect of the knee and ankle. An additional 11 markers were placed on the foot the subject self-reported as most affected by the neuropathy. A 3-segment foot model modified slightly from Bruening et al was used[29] (Figure 2). The less affected foot used 4 markers in a simple, single-segment marker set. The more affected foot was self-reported by the DPN subjects and randomly decided by the researchers for the controls. Subjects were brought to a carpeted walkway that lead up to the force plate (AMTI Inc, Watertown, MA, USA) with the pressure mat (Tekscan Inc, Boston, MA, USA) placed directly on top. The pressure mat/force plate combination was used to collect force and pressure data simultaneously for the same foot strike (Figure 3).

Subjects were first instructed to walk down the walkway at a natural, comfortable walking pace. Three trials were collected and used to determine each subject's self-selected walking speed. Next, the subjects walked down the walkway at a controlled speed of 1 m/s. This speed is midrange for subjects with DPN according to a review article by Allet et al[30] and is similar to DPN speeds found or used by other researchers.[20] A motor-driven pulley system was used to help subjects maintain the desired speed. This consisted of a string with small colored

flags that was attached at each end to a plastic wheel at waist height. The plastic wheel was calibrated to turn and pull the string at a consistent speed that the subjects could follow as they walked down the walkway (Figure 3). Subjects were allowed to practice walking with the device at the controlled speed until they were consistently matching it. Each subject's starting position was adjusted to ensure a full contact of a single foot on the pressure mat/force plate, allowing them to walk as naturally as possible with no targeting of foot strikes.

Data Analysis

A biomechanical model of the pelvis, thigh, shank, torso, head, and upper and lower arm segments was created according to common conventions while the 3-segment foot was made based off the model used by Bruening et al.[29] The model included a rearfoot, a mid/forefoot, and phalanges segments which were aligned with the subject's boney anatomy. Marker trajectories and force data were low-pass filtered at 6 Hz and 50 Hz, respectively, and joint angles were found based on an Euler angle rotation sequence (1-flexion/extension, 2-adduction/ abduction, 3-internal/external rotation). Joint kinematics and kinetics were calculated using Visual 3D software (C-Motion Inc, Germantown, MD, USA) for the stance phase of the controlled speed trials only. Only sagittal plane angles were used in this study. Midtarsal kinetics were only evaluated when the center of pressure passed anterior to the midtarsal joint.[31]

Whole curves were time normalized to 100% of stance and averaged across the 3 trials for each subject. Aggregate group means and standard error bands were then plotted for visualization. For statistical comparisons, mean between-group differences along with 95% confidence interval bands were plotted below each curve. Regions where these confidence interval bands separate from zero can be considered statistically significant at $\alpha = 0.05$. This is an approach that has been simplified from functional data analysis.[32] A few additional gait metrics

were extracted and compared between groups using independent t-tests (Table 3). These consisted of positive and negative work (integral of power) performed at the ankle and midtarsal joints.

Plantar pressure analysis consisted of metric comparisons. These consisted of the PPP and pressure-time integrals (PTI) for 7 different foot regions (hallux, lateral toes, medial forefoot, lateral forefoot, midfoot, medial heel, and lateral heel), the peak pressure gradients (PPG) for 3 regions (hallux, medial forefoot, and heel), and the PPP forefoot to rearfoot ratios for the whole foot. The 7 PPP and PTI regions were based on the default Tekscan analysis software, while the 3 PPG regions were manually created. All PP metrics were normalized by subject weight. Statistical comparisons were made using independent t-tests and a Benjamini-Hochberg procedure with a false discovery rate of 0.15. This procedure was used to account for the large number of t-tests performed in the analysis.

RESULTS

The diabetic and control groups were fairly well matched. Mass was on average 15 kg higher in the DPN group, but this was not significant. Any potential issues due to this were accounted for in the data processing by normalizing all PP and joint kinetics to weight. The mean Michigan Neuropathy Screening Instrument score was 8.9 out of 13, indicating moderate DPN progression. Seven out of 10 control subjects scored the minimum of 2, while 3 subjects indicated mild foot discomfort.

When walking at their self-selected walking pace, the diabetic subjects walked significantly slower than the control subjects (Table 2).[33] They increased speed 7.21% in order to match the controlled walking pace (1 m/s). Control subjects experienced an 18.74% decrease in

order to match the controlled speed. There was no significant difference found in the controlled speeds or time spent in stance between the 2 groups.

The DPN group had a higher dorsiflexion angle during terminal stance and during push off (Figure 4). The DPN group also exhibited larger positive and negative peak ankle power with a delay in transitioning from negative to positive power (Figure 4). Similar to the ankle power, the midtarsal power graph shows a delay in the transition from negative to positive power with a decrease in the peak positive power. Midtarsal positive work was significantly lower in DPN, though midtarsal negative work approached significance (Table 3). The DPN group's torso angle also showed a slight but not significant offset throughout stance with the diabetics staying more flexed by about 2° (Figure 5).

The hip angle graph shows an offset for almost the whole stance phase with the diabetic group staying in a more flexed position (Figure 6). There is a large delay seen in the transition from a hip extension to a hip flexion moment in the DPN group when compared to the control group. The DPN group also showed a decrease in peak hip flexion moment. Hip power was similar between groups with a mild decrease in power absorption in the DPN group (Figure 6). There was a slight delay in the transition from knee-flexion to knee-extension moment as well as knee power absorption (Figure 6).

The majority of the plantar pressure metrics were not found to be significant. The only exception to this was peak pressure under the lateral toes with the higher pressure exhibited by the control group (Table 4).

DISCUSSION

The overall purpose of this study was to assess full-body gait alterations in DPN at a controlled walking speed. We expected to see an increase in anterior trunk lean, a decrease in

ankle and midfoot power generation and an increase in knee and hip power generation. Our hypotheses were only partially supported by our findings. Our results showed multiple gait variations with only 1 (out of 18) PP alteration when walking speed was controlled.

Gait Alterations

Early Stance (initial contact through loading response). We saw very few gait differences between groups in early stance. The knee and hip angle graphs show slight increases in flexion at initial contact in individuals with DPN, but these are minor and not statistically significant. Previous studies have shown that many individuals with DPN have a more cautious style of gait,[26] which results in increased lower limb flexion at initial contact. It is possible that our subjects, who were in the initial stages of neuropathy, were not sufficiently disabled to have this compensation strategy.

Interestingly, the diabetic group had a significant increase in MTP extension at initial contact and through loading response (Figure 4). There seem to be 2 possible explanations for this increase. First, the diabetic subjects could have weakness or lack of motor control of the dorsiflexors,[24] which eccentrically control the rate of plantar flexion during loading response. Weakness of the tibialis anterior may require excessive recruitment of other dorsiflexors, such as the toe extensors. The second possible explanation is an inability to relax the extensor hallucis longus muscle from swing due to decreased motor control related to DPN. Additional analysis of swing phase as well as inclusion electromyography may elucidate this mechanism.

Midstance (midstance to terminal stance). Throughout midstance, the hip appears to compensate for distal muscle weakness. This is more apparent in the hip moment than in the knee moment and knee/ hip powers (Figure 6). The hip extension moment is increased and prolonged in individuals with DPN. This difference could be due to an anterior trunk lean and/or

compensation for weakness in the posterior tibial muscles. Increased anterior trunk lean could result in passive stretch of the hip extensor muscles, increasing tension. Although torso angle was not found to be significantly different (Figure 6), there appeared to be a slight anterior offset of approximately 2°, but with a large variability. Further research on the combination of anterior pelvic tilt and anterior trunk lean with lordosis may clarify these results. Alternatively, the altered hip moment may be a compensatory mechanism for ankle muscle weakness. Due to reduced momentum from a weak contralateral push off (see Late Stance), the hip extensor muscles are needed to pull the body forward, manifesting in an increased and prolonged hip extensor moment.[26]

Once the body passes anterior to the ankle, the plantar flexor muscles are needed to control anterior tibial progression.[34] The dorsiflexion angle graph shows a slight increase in angular velocity and peak dorsiflexion in early terminal stance, which suggests a minor collapse at the ankle joint instead of a controlled roll forward.[34,35] This also is a sign of weak plantar flexors.[26,36] The increase in peak dorsiflexion angle could lead to added stress on the soft tissue around the joint (ie, power absorption). This helps explain the increase seen in the peak negative ankle power. DiLiberto et al[22] also found an increase in negative ankle power and work, theorizing that this imbalance in power during the common task of walking could lead to midfoot pathologies.

Late Stance (terminal stance through toe off). The push-off phase of stance in the diabetic group suggests ankle and foot weakness along with possible sensation loss.[24,37] In late stance we saw delays in push-off as evidenced by later transitions in the hip joint moment, ankle angle, and ankle and midtarsal powers. These could be explained by a need for individuals with DPN to reach sufficient tissue stretch to engage compromised proprioceptive mechanisms. Despite the

delayed ankle power generation, the positive work done at the ankle was not different between the groups. This could be due to increased passive stretch and subsequent energy return of the Achilles tendon. Alternatively, if this energy is being dissipated rather than stored, the late increase in ankle power may indicate a quick delayed burst of muscle activity indicating developing motor control issues. Either explanation points toward a weakness in ankle plantar flexors.

Midtarsal joint mechanics show similarities to the ankle through early terminal stance, including increased negative power. Surprisingly, there was not a concomitant increase in midtarsal dorsiflexion with increased ankle dorsiflexion. Midtarsal power generation and positive work done were also reduced.[22] This may be due to weakness of the intrinsic foot muscles[11,37] and/or decreased engagement of the windlass mechanism. The latter is apparent in the delayed onset of MTP extension and reduced peak angle, suggesting reduced passive power transfer from the MTP joint to the midtarsal joint.[38]

Plantar Pressure

Plantar pressure was less informative than joint kinetics in identifying gait deviations in individuals with DPN. The only significant PP difference found was a decrease in PPP under the lateral toes in late stance. We speculate that this may be due to premature activity of the toe extensors in preparation for swing. However, this would have to be confirmed by EMG. The lack of additional significant findings may be related to our specific subject pool, which were in the initial stages of DPN. However, the lack of significant PP differences in this study contrasted with the clear kinetic differences suggest that PP is a less informative measure for DPN gait alterations.

Limitations and Future Research

There were several limitations in this study. First, only the stance phase of gait was evaluated. We chose to exclude swing and focus on stance due to the possible clinical connection to ulcer and other pathology development. In this study we used a controlled speed which may have induced additional compensations in individuals with DPN. The chosen controlled speed of 1 m/s required individuals with DPN to speed up while requiring the control subjects to slow down. We felt that this minimized compensations in both groups and allowed us to best isolate the effects of DPN from walking speed. Our DPN participants were not as involved as some previous studies. A more involved subject sample may have resulted in additional significant differences. However, a greater degree of involvement may have had more difficulty in meeting the task demands. We specifically targeted subjects in the initial stages of neuropathy in order to avoid the numerous confounding effects of foot deformities and injuries, attempting to isolate only the effects of DPN. Last, we attempted to match the demographics of the two groups, however the age range in the DPN group was larger and the mean mass was slightly higher. A better age match may have increased statistical power. Kinetic results were normalized to body mass minimizing any confounding effects.

Future research could focus on longitudinal studies or on controlled walking speeds in samples of a varying disease progression. This could help expand the picture of how gait mechanics change with DPN over time and emphasize the changes that have already been seen. Many of our findings were related to a possible decrease in foot and ankle strength. Future studies could include a strength test that looks at both strength over a short period of time (ie, 1 rep max) and over a long period of time (endurance strength) in order to better understand the

relationship between the ankle muscle strength, fatigue and the increased peak power shown in this study.

Conclusion

This study demonstrated the need to control speed to better isolate the effects of DPN on walking gait. Plantar pressure measures were similar between groups with the exception of a decrease in lateral toe PPP. The gait compensations demonstrated by the DPN group seemed to be consistent with distal muscle weakness. This is shown by the increased dorsiflexion angle and increased negative power at the ankle and midtarsal joints and the compensatory prolonged hip extension moment. It is also apparent in the delays in ankle angle and power, midtarsal power, MTP angle and hip power graphs. By using a multisegment foot model in conjunction with a traditional full-body model, we were able to identify proximal compensations for distal weakness. The results may be useful in assessing DPN gait progression and developing intervention protocols.

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Table 1 Demographics Compared Between DPN and Control Groups

	DPN Group	Control group	p value
Height (cm)	178.79 ± 8.55	180.00 ± 6.37	0.723
Mass (kg)	108.78 ± 16.67	92.87 ± 14.50	0.035
Age (yrs)	61.50 ± 13.53	59.40 ± 7.50	0.673
MNSI score	8.90 ± 2.47	2.40 ± 0.70	< 0.001*

* Indicates a significant difference between groups

Table 2 Spatiotemporal Variables Compared Between DPN and Control Groups

	DPN Group	Control Group	p value
Self-selected Speed (m/s)	0.91 ± 0.15	1.21 ± 0.14	< 0.001*
Controlled Speed (m/s)	0.99 ± 0.09	1.02 ± 0.06	0.308
Stance Time (s)	0.71 ± 0.06	0.73 ± 0.02	0.242

* Indicates a significant difference between groups

Table 3 Gait Metrics Compared Between DPN and Control Groups

	DPN Group	Control Group	p Value
Midtarsal Positive Work	0.08 ± 0.01	0.11 ± 0.02	0.005*
Midtarsal Negative Work	−0.04 ± 0.02	−0.02 ± 0.01	0.066
Ankle Positive Work	0.13 ± 0.05	0.11 ± 0.03	0.451
Ankle Negative Work	−0.16 ± 0.17	−0.09 ± 0.05	0.212

* Indicates a significant difference between groups

Table 4 Plantar Pressure Metrics Compared Between DPN and Control Groups

	DPN Group	Control Group	p Value
Lateral Toes PPP	1.49 ± 0.97	2.59 ± 0.65	0.011*
Medial Forefoot PPP	5.16 ± 1.83	5.13 ± 1.50	0.967
Medial Heel PPP	3.49 ± 0.77	4.34 ± 1.02	0.054
Lateral Heel PPP	3.17 ± 0.59	3.91 ± 1.01	0.064
Hallux PTI	0.52 ± 0.25	0.29 ± 0.17	0.028
Medial Forefoot PTI	0.74 ± 0.18	0.76 ± 0.28	0.841
Hallux PPG	5.96 ± 1.31	5.20 ± 3.47	0.543
Medial forefoot PPG	4.41 ± 1.68	4.77 ± 1.80	0.655
Heel PPG	4.05 ± 3.21	3.13 ± 0.80	0.390
FF/RF Ratio	1.15 ± 0.43	1.41 ± 0.40	0.195

* Indicates a significant difference between groups

Figure 1 Full-Body Marker Set

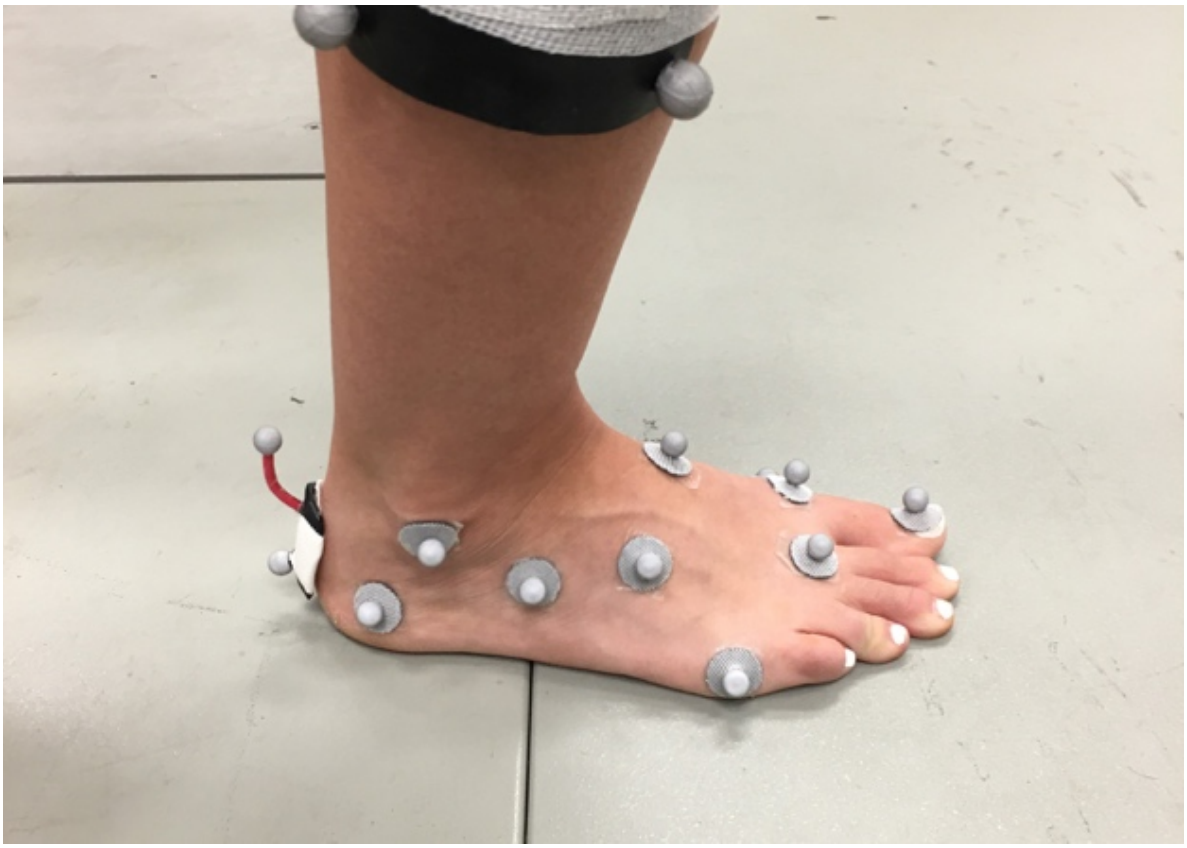

Figure 2 Example of Modified Foot-Marker Set

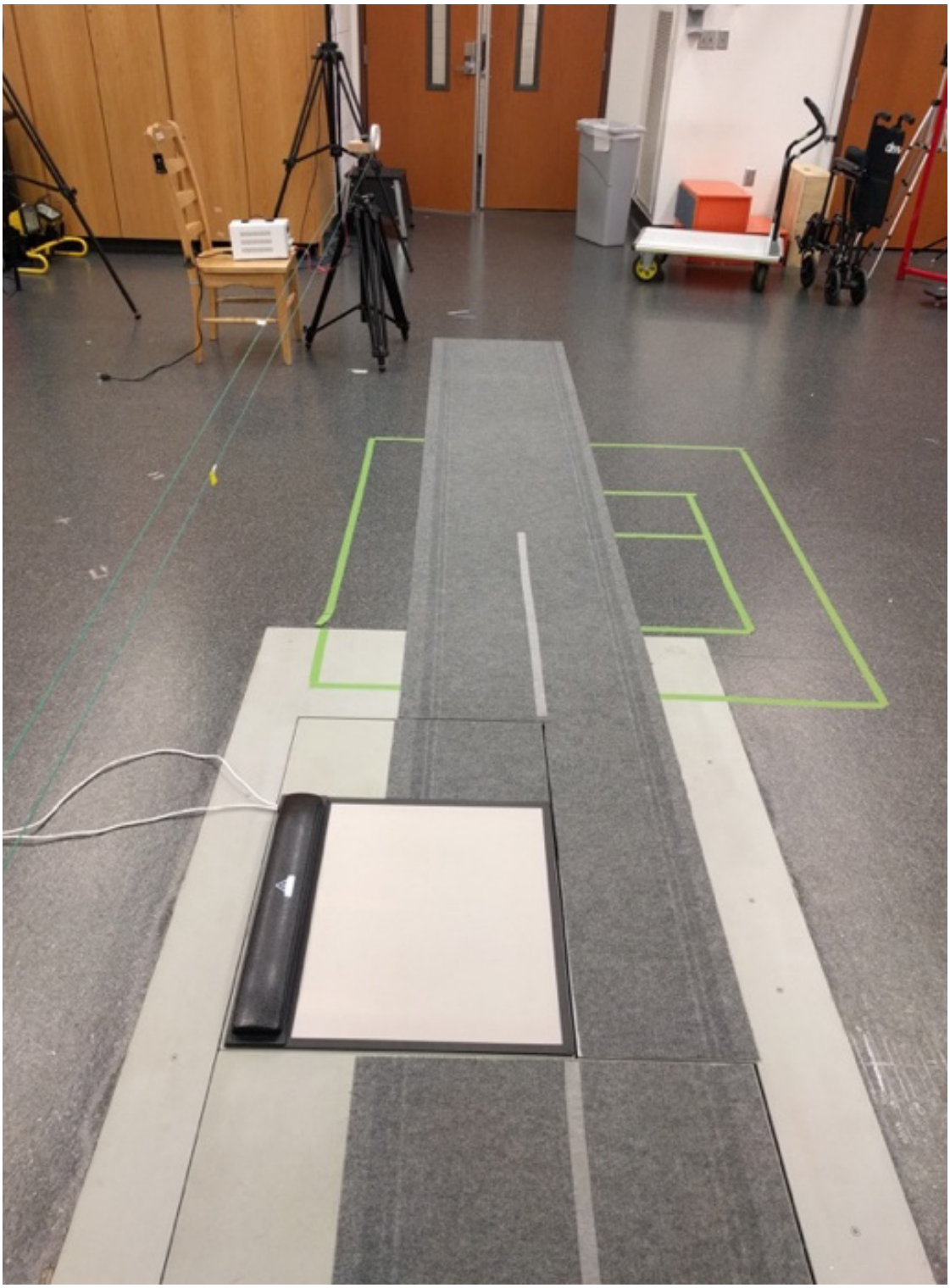

Figure 3 Motorized Speed-Control String with Small Colored Flags on Left of Walkway

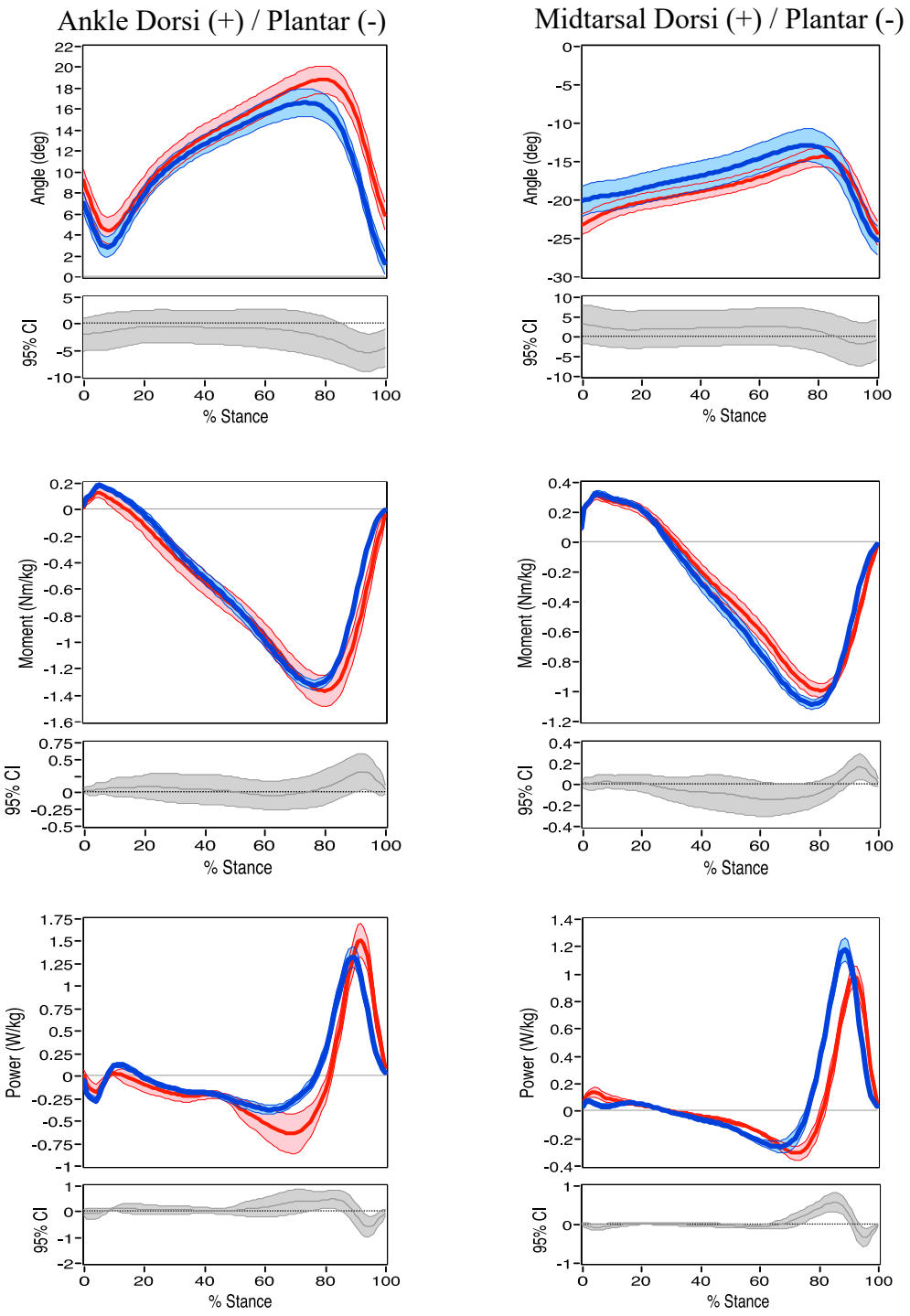

Figure 4 Ankle and Midtarsal Angles, Moments, and Power Top row: Ankle and midtarsal sagittal angles. Middle row : Ankle and midtarsal moments. Bottom row: Ankle and midtarsal power. Red curves represent the DPN group, blue represent control group. Solid lines are the group mean with shaded areas representing standard error bands. The 95% confidence interval is plotted below to show statistical significance where it departs from 0.

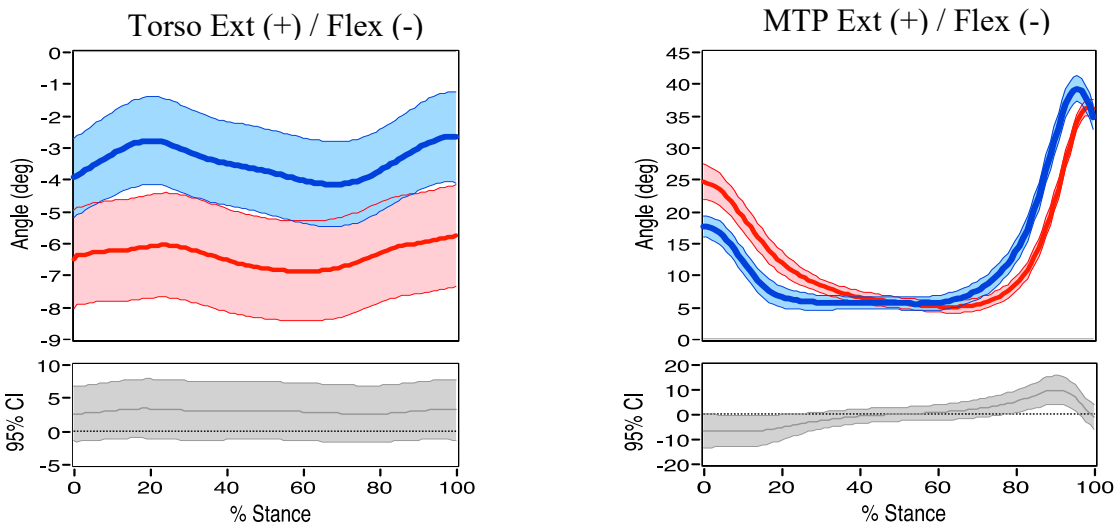

Figure 5 Torso Sagittal Angle and MTP Sagittal Angle. Red curves represent the DPN group, blue represent the control group. Solid lines are the group mean with shaded areas representing standard error bands. The 95% confidence interval is plotted below to show statistical significance where it departs from 0.

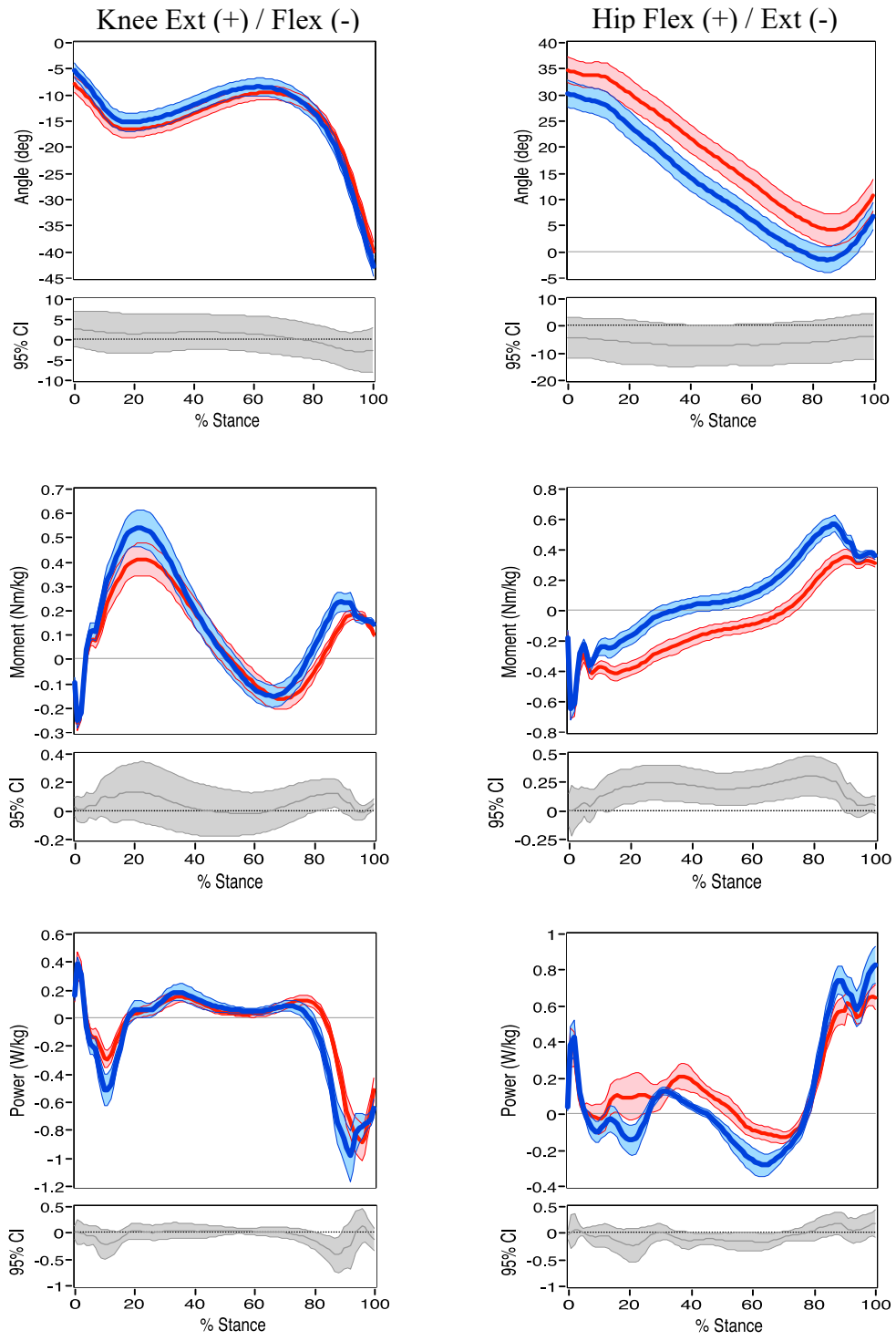

Figure 6 Knee and Hip Angles, Moments, and Power Top row: Knee and hip sagittal angles. Middle row: Knee and hip moments. Bottom row: Knee and hip power. Red curves represent DPN group, blue represent control group. Solid lines are group mean with shaded areas representing standard error bands. The 95% confidence interval is plotted below to show statistical significance where it departs from 0.