The Effect of Stiff Foot Plate Length on Walking Gait Mechanics

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Exoskeletons are increasingly being used to treat gait pathologies. Many of these exoskeletons use a foot plate to actuate the foot, altering the effective stiffness of the foot. Stiffness of the biological foot and ankle play an important role in the energy modulating function of the leg, so it is important to examine how a foot plate in and of itself impacts gait. Therefore, this study quantified how foot plates themselves alter the walking gait of 16 healthy young adults. The effect of foot plate length was also examined through the use of two foot plates, one that ended at the metatarsals and one that extended past the toes, about 20% longer. Gait parameters examined included walking speed, step frequency, joint angles for the hip, knee, ankle, forefoot, and toe, ground reaction forces (GRF), and foot-ankle power. The most significant changes were caused by the full plate, which caused an average 13% decrease in ankle range of motion (ROM) and a 23% decrease in forward GRF at push off. The shorter plate also decreased ankle ROM to a lesser degree. This indicates that the presence of a foot plate impacted foot and ankle kinematics. However, the presence of the tested foot plate had no effect on walking speed or hip or knee kinematics. This indicates that subjects were mostly able to compensate both kinematically and energetically via their foot and ankle for the increased foot stiffness due to the tested foot plate.

1 Introduction

Exoskeletons may offer advantages over traditional gait rehabilitation for many patient populations [1–6]. Often, exoskeletons use a foot plate to actuate the foot of the wearer [7–11]. There is no consensus on the best design for these foot plates. Some extend past the toes, and some only cup the heel. However, foot plates may impact user gait, and these unintended effects could inhibit the usefulness of the exoskeleton.

An important foot function during walking is energy modulation. The foot dissipates some energy at heel strike [12, 13]. In the second half of stance, the toes dorsiflex and engage the windlass mechanism, which raises the arch of the foot and passively stores energy in the plantar aponeurosis (PA) [14–16]. This energy is released in conjunction with the contraction of the plantar intrinsic muscles under the foot to generate some of the push off force [17, 18]. This force helps to redirect the body center of mass (COM) and allows continued walking [19].

In prosthesis studies, altering foot and ankle stiffness alters gait. Changing overall foot or ankle stiffness affects the energy absorbed and produced by the foot during stance, which affects the energy of the whole body and changes overall power use [20–22]. However, the overall metabolic cost of walking remains relatively constant [23]. Kinematically, it usually only affects the ankle [24, 25]. Changing toe joint stiffness impacts gait mechanics as much as changing ankle joint stiffness [25]. Therefore, it is possible that altering only foot stiffness will affect gait mechanics, but this has not yet been systematically investigated. However, changing prosthetic foot stiffness does not generally affect ground reaction forces (GRF) [26], suggesting that humans may compensate for changes in foot or ankle stiffness to maintain a desired push off force.

Most research in this area was done with prosthetic feet, although some stroke research showed that changing the length of an Ankle-Foot Orthosis (AFO) foot plate, in conjunction with other design parameters, altered gait mechanics [27]. Shoes may also alter foot stiffness, but when compared with shoe stiffness, the bending stiffness of the forefoot overshadows the stiffness of the shoe [28]. Because the shoe is in parallel with the foot, the change

(a) (b) (c)

Fig. 1: (a) Shoes worn by the subjects for first part of study. (b) Full length foot plate. (c) Three-quarter length foot plate. Foot plates shown upside down.

in overall stiffness is likely negligible. Nevertheless, shoes may cause kinematic changes to the toe specifically [29, 30], although this is difficult to measure [31]. Compared to barefoot walking, wearing shoes increases both knee range of motion (ROM) and ankle plantarflexion at push off [30]. In contrast, increasing ankle-foot stiffness using an AFO or prosthesis leads to a consistent decrease in ankle ROM and an occasional increase in stance knee flexion [23, 25, 26, 32], so it is not clear how a foot plate will alter joint kinematics. Shoes also increase stride length, decrease step frequency, and increase GRF in the vertical direction at heel strike compared with barefoot walking [30]. However, it is not clear how shoes alter the anterior/posterior (A/P) GRF at push off [30]. Stiff plates in shoes increase sprint performance, suggesting an optimal shoe stiffness although it is unclear where the benefits arise from [33].

Since changing foot stiffness affects the entire body [20–22], and the windlass mechanism is an integral part of generating stiffness [14–16], it may be important to preserve the normal windlass mechanism function during rehabilitation. In turn, this should influence how an exoskeleton foot plate is designed. Because the foot plate is in parallel with the foot and has a significant bending stiffness relative to the foot (order $10^2 N/m$), the overall bending stiffness of the foot plus foot plate system increases. For conciseness, we will refer to the foot plus foot plate system as the foot. We hypothesize that a moderate stiffness foot plate will affect spatiotemporal parameters, and hip, knee, ankle and foot kinematics. Additionally, we hypothesize that the effect will be more pronounced with a longer foot plate that extends past the toes. This is because a longer foot plate restricts the toe joint, thereby forcing the foot to act as a single stiff body (effective toe joint stiffness increased by order $10^2 N/m$) rather than two links connected with a compliant joint (no increase in toe joint stiffness). This effectively makes the foot stiffer. To investigate these hypotheses, healthy subjects were tested with two different lengths of foot plates while walking. The differences in kinematics and kinetics for walking with and without a foot plate were quantified. In order to effectively investigate foot kinematics, a ballet slipper was used during some experiments to hold the foot plate on the foot and allow the use of the Oxford foot model, which captures the motion of the hindfoot, forefoot, and toe [34].

2 Methods

There were two parts to this study. The first part, referred to as part one, examined twelve subjects wearing foot plates in shoes, which is how foot plates are normally utilized. The Plug-in Gait (PiG) model was used, which assumes a rigid foot [35]. This part of the study was used to investigate both hypotheses except for the foot kinematics in the first hypothesis. However, the foot itself deforms while walking [36], and initial results indicated that measuring changes in foot segment movement may be important. Therefore, in order to test the hypothesis that a foot plate alters foot kinematics, four more subjects were tested using the Oxford foot model [34]. This part of the study was used to investigate both hypotheses. Initial results also suggested that the foot plates stored energy due to elastic deformation, which could alter the energy flow. Therefore, for the second part of the study, referred to as part two, the elastic energy in the foot plate over the stance period was calculated using finite element analysis (FEA) and compared to the total energy of the ankle, foot, and foot plate calculated using a unified deformable (UD) model [37].

2.1 Foot Plates

The foot plates were made from 5 mm delrin with a thin 2 mm layer of foam on top for comfort. The foot plates replaced the insole of the provided shoes (Adidas, Fig. 1a). Two versions were tested: a foot plate that extended past the toes (Fig. 1b) called the full plate, and a foot plate that ended at the metatarsal joint (Fig. 1c) called the three-quarter plate; this plate was about 20% shorter, and had a 5 mm pad of denser foam under the toes to maintain a constant height. For part two, in order to measure the deformation and calculate the elastic energy

Fig. 2: Modified ballet slipper for second part of study, foot markers, and foot plate markers (three-quarter plate shown).

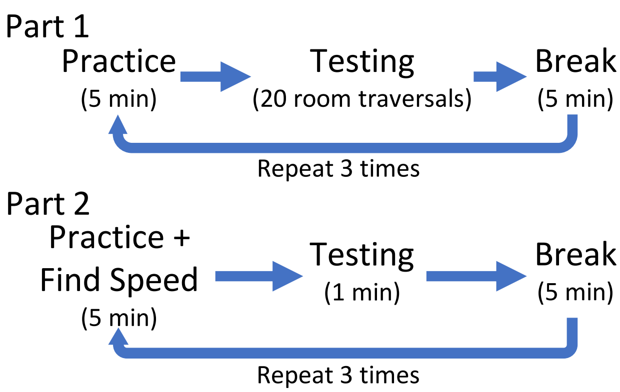

Fig. 3: Procedure for each part of the study. This sequence is repeated once for each condition in a randomized order.

in the foot plate, reflective markers were attached to the foot plates using set screws (Fig. 2). For the full plate, fifteen markers were used, eight on the lateral side and seven on the medial. For the three-quarter plate, eleven markers were used, six on the lateral side and five on the medial.

2.2 Experimental Protocol

Twelve healthy adult subjects (6 male, 6 female, age 20 to 30 years, mass 70.8 kg ± 15.4 kg, height 1.68 m ± 0.07 m) participated in part one. Four healthy adult subjects (2 male, 2 female, age 20 to 30 years, mass 75.5 kg ± 13.1 kg, height 1.70 m ± 0.06 m) participated in part two. An IRB protocol was approved, and informed consent was collected from each subject. Subjects for part one wore a pair of provided mens shoes sized 6, 7, 8, 9, 10, or 11 (Fig. 1a). To fit the modified foot plate for part two, those subjects fit mens size 9 shoes.

For part one, subjects walked overground (Fig. 3). The PiG model [35] was used. Heel and toe markers were attached to the outside of the shoe, with the rest attached to the subjects' skin. Three conditions were tested in random order: 1) a control condition with the manufacturer's insole, 2) a full plate condition with the insole replaced by the full plate, and 3) a three-quarter plate condition with the insole replaced by the three-quarter plate. Prior to data collection, subjects adjusted to walking with the foot plates (approximately 5 minutes), and an appropriate starting position was found. Each subject completed at least 20 trials per condition. A trial consisted of walking approximately 6 meters, stepping with each foot on each force plate. Kinematic data were collected at 100 Hz (Vicon, Oxford, UK) and kinetic data were collected at 1000 Hz using force plates centered along the walkway (Bertec, Columbus, OH).

For part two, subjects walked on an instrumented, split-belt treadmill with force plates under each foot (Bertec, Columbus, OH) wearing ballet slippers. Both the Oxford foot and PiG model were used (Fig. 2). While foot plates are typically used in shoes, using shoe mounted markers to estimate physiological foot kinematics is unreliable [38]. Cutting holes in shoes compromises their structure, changing the effect they have on gait, while still not guaranteeing accurate kinematics [31]. Instead, ballet slippers were used because they could withstand having holes cut into them to allow accurate marker placement directly on the bony landmarks of the foot. Ballet slippers approximate barefoot walking, meaning the control condition for part two can be considered unshod. The same three conditions as part one were tested in random order (Fig. 3). The preferred speed for each condition was determined by gradually increasing treadmill speed until it was too fast, then decreasing it until it was too slow. This was performed a total of three times, and the average of all six values was used for the experimental trial [39]. During this time, the subjects adjusted to walking with the foot plates (approximately 5 minutes). For each trial, subjects walked for one minute while kinematic and kinetic data were recorded. Because it was not possible to remove the ballet slippers without removing markers, the foot markers were removed, replaced, and recalibrated between each condition. Replacing markers between trials may cause an absolute shift in the overall foot intersegmental angles, but does not affect the shape of the motion or the ROM [34]. Subjects also wore a pressure sensing insole attached by a cord to the computer; these data were not analyzed in this paper.

2.3 Data Processing

Data for part one were split into strides for each leg; a stride goes from heel strike to ipsilateral heel strike. For part one, gait events were determined by peaks in the heel marker data and verified visually for each stride. For part two, GRF was used to identify gait events based on 1% of subject weight. Strides with large gaps in the data (greater than 20 frames), with incorrectly identified gait events, or that had hip, knee, or ankle joint ROM outside of 3 standard deviations of the mean were removed as outliers. In addition, strides in which the subject did not step cleanly on the force plate were discarded.

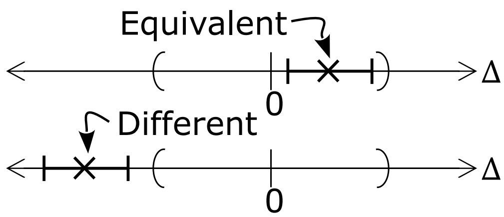

Fig. 4: Example of statistical equivalence and difference. Parentheses represent equivalence bounds, x's represent the mean difference of a parameter with standard deviation bars. If the standard deviation of the tested parameter is inside the bounds, it is statistically equivalent. If it is outside the bounds, it is statistically different. Otherwise, it is inconclusive.

For part one, spatiotemporal parameters (walking speed and stride frequency), ROM for the hip, knee, and ankle in all three planes, peak knee flexion in the first half of stance, GRF peaks in all three planes, and work done by the foot/foot plate were analyzed for statistical differences and equivalence. For part two, the above parameters as well as ROM of the toe with respect to the forefoot, the forefoot with respect to the hindfoot, and the hindfoot with respect to the tibia (closest to PiG ankle angle), were calculated in three dimensions. Only strides on the force plates were used for kinetic measurements; all strides were included for the other parameters. For part one, approximately one third of the strides had GRF data, while GRF data was available for every step in part two. Average ROM was calculated by first finding the ROM for each step and then calculating the mean over all recorded steps. GRF data was split into the first and second half of stance and the maximum absolute value was taken as the peak. GRF data were normalized by subject mass.

The PiG model assumes a rigid foot and calculates ankle power accordingly. However, the foot is not rigid. Instead, the existing, validated unified deformable (UD) model was used to calculate ankle-foot power [37, 40, 41]. This model treats all structures below a known rigid structure, such as the shank, as one deformable object. Because this model only captures power transmitted from the deformable object, it neglects the mass of the object and does not calculate power due to deformation within the deformable object. Power is given by

$$P = F_{GRF} \bullet V_{cop} + M_{Ext} \bullet \omega_{shank} \tag{1}$$

where P is the power transmitted to the shank, F_{GRF} is GRF, M_{Ext} is the external moment applied to the shank, and ω_{shank} is the angular velocity of the shank. V_{cop} is the velocity of center of pressure under the foot, calculated by

$$V_{cop} = V_{com-shank} + (\omega_{shank} \times r_{com-cop}) \tag{2}$$

where $V_{com-shank}$ is the velocity of the shank COM relative to the walking surface. $r_{com-cop}$ is the distance from the center of pressure to the shank COM. UD power was normalized by subject mass and compared qualitatively between conditions. Power was integrated to find the total work of the UD segment over the stride, which was statistically compared across conditions. Some subjects from part two had significantly higher peaks in UD power than is typical for overground walking. However, trends for each subject were consistent.

To estimate energy storage, the foot plates were modeled in Abaqus, and the measured deflection data from the foot plate markers were used to deform them. To obtain deflection data, the foot plate marker data were split into strides, and outliers were removed using the same process as before. The vertical marker positions were re-sampled so that each stance period had 600 frames. The position of each marker was then averaged across all stance periods for each frame. This deflection data was down sampled to 31 equally-spaced frames plus 9 additional frames near toe-off to capture the quickly changing deflection. The deflection data was manually input into Abaqus. Within Abaqus, the location of the specified deflection points were defined using face and cell partitions and matched to the physical marker positions as closely as possible. However, there were small errors that resulted in small, but non-zero, estimates of elastic energy even for an undeformed foot plate. The sole of the plate was allowed to deform, but the majority of the deformation occurred in the sagittal plane. The elastic energy was calculated using the experimental deflection data, assuming an elastic modulus of 2410 MPa [42] for the foot plates. This elastic energy was normalized by subject mass and qualitatively compared to the UD segment.

2.4 Statistical Testing

In all cases, the three-quarter and full plate conditions were compared to the control condition. Statistical differences and equivalences were found using standard t-tests and two one-sided tests (TOST) [43], respectively ($\alpha = 0.05$). A statistical difference indicates that observed differences were not due to measurement noise, while statistical equivalence indicates that changes were not meaningful (Fig. 4). For example, a change in walking speed of 0.02 m/s is inconsequential regardless of whether or not it is a statistically significant difference. This means it was possible for a difference to be both statistically different and equivalent; when this occurred, the difference was reported as equivalent. If the difference was neither statistically different nor equivalent, the difference was reported as inconclusive. Data were considered equivalent if they were within equivalence bounds of the control condition. For most comparisons, these bounds were 10% of the mean value of the control condition. For joint ROM, a Cohen's d value of 1 defined the equivalence bounds. These equivalence bounds corresponded to approximately $\pm 0.1 m/s$ for

Table 1: Results from part one. Mean and standard deviation of the difference between foot plate conditions and control over a stride. Dir. indicates if the foot plate significantly increased (↑) or decreased (↓) the parameter, if the foot plate condition was equivalent (=), or if it was inconclusive (X). Num. indicates the number of subjects that exhibit the difference, were equivalent, or were inconclusive out of twelve subjects. If Dir is =, statistical equivalences are reported ($* = p < 0.05$, $** = p < 0.005$, $*** = p < 0.001$). If Dir is ↑ or ↓, statistical differences are reported ($\dagger = p < 0.05$, $\dagger\dagger = p < 0.005$, $\dagger\dagger\dagger = p < 0.001$)

Parameter	Three-Quarter				Full			
	Mean	Std. Dev.	Dir.	Num.	Mean	Std. Dev.	Dir.	Num.
Walking Speed (m/s)	0.02***	0.11	=	12	0.004***	0.12	=	12
Step Frequency (Hz)	0.00***	0.06	=	12	-0.00***	0.06	=	12
Hip ROM PF/DF (°)	0.29***	3.39	=	11	0.07***	3.40	=	8
Hip ROM AD/AB (°)	0.02***	2.06	=	7	-0.49***	2.07	=	6
Hip ROM INV/EV (°)	-0.19***	12.3	=	10	-1.80***	12.5	=	8
Knee ROM PF/DF (°)	-0.80***	11.6	=	12	-0.56***	11.4	=	10
Knee ROM AD/AB (°)	1.09***	13.1	=	11	0.00***	12.7	=	7
Knee ROM INV/EV (°)	-0.91***	23.1	=	12	-0.77***	22.9	=	10
Early Knee Flexion (°)	0.57***	3.01	=	8	-0.56**	2.73	=	6
Ankle ROM PF/DF (°)	-3.74***	29.9	=	6	$-4.96^{\dagger\dagger}$	30.3	↓	8
Ankle ROM AD/AB (°)	-1.03***	10.4	=	8	-0.95***	10.17	=	10
Ankle ROM INV/EV (°)	-0.89***	19.4	=	11	1.42**	20.3	=	11
Braking Vert. (N/kg)	0.03***	0.74	=	12	-0.45***	0.77	=	12
Push Off Vert. (N/kg)	-0.10***	0.65	=	12	-0.34***	0.67	=	12
Braking A/P (N/kg)	0.04***	0.29	=	9	0.01***	0.30	=	6
Push Off A/P (N/kg)	0.01***	0.30	=	9	$-0.34^{\dagger\dagger\dagger}$	0.29	↓	11
Braking M/L (N/kg)	0.02***	0.20	=	7	-0.01	0.19	X	3
Push Off M/L (N/kg)	0.04	0.19	X	4	$0.03^{\dagger}$	0.19	↑	7
UD Work (J/kg)	0.12	0.61	X	10	0.00	0.62	X	10

speed, $\pm 0.1 Hz$ for stride frequency, $\pm 5°$ for ROM, $\pm 2°$ for intersegmental ROM, $\pm 1 N/kg$ for vertical GRF, and $\pm 0.5 N/kg$ for horizontal GRF. Because subjects differed in their response to the foot plate, both the average response over all subjects and the number of subjects exhibiting the typical response were reported. Note that for part two, per-subject speed changes were defined as changes in treadmill speed and did not require statistical testing.

3 Results

3.1 Spatiotemporal Parameters

In part one (N=12), average walking speed was 1.23 m/s and average stride frequency was 0.89 Hz. In part two (N=4), average walking speed was 1.13 m/s and average step frequency was 0.95 Hz. For both foot plate conditions in both parts of the study, walking speed and stride frequency were equivalent (Tab. 1,2).

3.2 Joint Kinematics

For both parts of the study, hip and knee ROM over the entire stride in all three direction were statistically equivalent between conditions (Fig. 5, Tab. 1,2). With the exception of stance knee flexion, hip and knee kinematics appeared to be equivalent for all three conditions. Stance knee flexion was not increased in most subjects. However, two subjects in part one (N=12), and one from part two (N=4), increased stance knee flexion with a three-quarter plate. Additionally, four subjects in part one and one from part two, increased stance knee flexion with a full plate. For these subjects, stance knee flexion increased from 5.8° (control) to 9.1° (full plate). In contrast, two subjects from part one decreased stance knee flexion with

Table 2: Results from part two. Mean and standard deviation of the difference between foot plate conditions and control over a stride. Dir. indicates if the foot plate significantly increased (↑) or decreased (↓) the parameter, if the foot plate condition was equivalent (=), or if it was inconclusive (X). Num. indicates the number of subjects that exhibit the difference, were equivalent, or were inconclusive out of four subjects. Reported walking speed refers to the speed of the treadmill. If Dir is =, statistical equivalences are reported ($* = p < 0.05$, $** = p < 0.005$, $*** = p < 0.001$). If Dir is ↑ or ↓, statistical differences are reported ($\dagger = p < 0.05$, $\dagger\dagger = p < 0.005$, $\dagger\dagger\dagger = p < 0.001$)

Parameter	Three-Quarter				Full			
	Mean	Std. Dev.	Dir.	Num.	Mean	Std. Dev.	Dir.	Num.
Walking Speed (m/s)	-0.05***	0.04	=	4	-0.08***	0.08	=	3
Step Frequency (Hz)	-0.02***	0.00	=	4	-0.04***	0.05	=	4
Hip ROM PF/DF (°)	-1.49***	4.44	=	4	-1.53***	4.60	=	3
Knee ROM PF/DF (°)	0.84***	7.88	=	2	2.82***	7.12	=	2
Early Knee Flexion (°)	1.56***	12.5	=	3	4.60***	11.9	=	2
Ankle ROM PF/DF (°)	$-5.95^{\dagger\dagger\dagger}$	9.78	↓	3	$-11.6^{\dagger\dagger\dagger}$	8.71	↓	3
Hind ROM PF/DF (°)	$-5.60^{\dagger\dagger\dagger}$	9.21	↓	2	$-11.17^{\dagger\dagger\dagger}$	8.20	↓	4
Hind ROM AB/AD (°)	$0.92^{\dagger}$	5.30	↑	3	-0.14***	5.40	=	2
Hind ROM INV/EV (°)	-0.61***	4.34	=	2	-1.35***	4.75	=	2
Fore ROM PF/DF (°)	$-3.03^{\dagger\dagger\dagger}$	4.98	↓	3	$-3.19^{\dagger\dagger\dagger}$	4.09	↓	2
Fore ROM AB/AD (°)	-2.28***	4.03	=	2	$-5.44^{\dagger\dagger\dagger}$	3.48	↓	3
Fore ROM INV/EV (°)	1.71	12.6	↓	3	$-2.91^{\dagger\dagger\dagger}$	3.68	↓	3
Toe ROM PF/DF (°)	$-4.22^{\dagger\dagger\dagger}$	9.02	↓	2	$-11.4^{\dagger\dagger\dagger}$	7.93	↓	4
Toe ROM AB/AD (°)	0.78***	4.67	=	1	$-2.78^{\dagger\dagger\dagger}$	3.30	↓	4
Toe ROM INV/EV (°)	$-2.47^{\dagger\dagger\dagger}$	7.57	↓	2	$-5.38^{\dagger\dagger\dagger}$	7.00	↓	2
Braking Vert. (N/kg)	-0.44***	1.16	=	4	-0.63***	1.21	=	4
Push Off Vert. (N/kg)	-0.32***	0.83	=	4	-0.26***	0.86	=	4
Braking A/P (N/kg)	-0.12*	0.64	=	3	-0.01	0.75	X	0
Push Off A/P (N/kg)	-0.22***	0.58	=	2	$-0.49^{\dagger\dagger\dagger}$	0.56	↓	3
Braking M/L (N/kg)	-0.07**	0.32	=	1	$-0.14^{\dagger\dagger\dagger}$	0.31	↓	3
Push Off M/L (N/kg)	-0.02***	0.23	=	3	$-0.04^{\dagger\dagger\dagger}$	0.22	↓	3
UD Work (J/kg)	-0.23	1.48	X	1	-0.09	1.14	X	2

a foot plate indicating subject-specific differences in knee compensation.

In part one (N=12), the three-quarter plate did not significantly affect sagittal ankle ROM in six subjects but decreased it in five subjects (Fig. 5, Tab. 1). With the full plate, eight subjects decreased ankle ROM by approximately 5°, an 11% decrease. Ankle ROM was equivalent for the remaining subjects. In part two (N=4), both plates decreased ankle ROM for the same three subjects. In these subjects, the three quarter plate reduced ankle ROM by about 5.9° (18% decrease) while the full plate reduced ankle ROM by almost twice as much (11.9° or a 35% change). Despite these changes in ROM, the timing of the peaks and the overall shape of the angle curve remained similar for all conditions. In both parts of the study, ankle kinematics in the other two directions were equivalent across conditions.

In part two (N=4), the hindfoot experienced significant decreases in plantarflexion (PF/DF) ROM for both foot plate conditions (Fig. 6, Tab. 2) Hindfoot PF/DF ROM decreased from 29.1° to 23.5° (19% change) with a three-quarter plate and further to 17.9° (38% change) with a full plate. The three-quarter plate increased hindfoot abduction (AB/AD) ROM by 26%

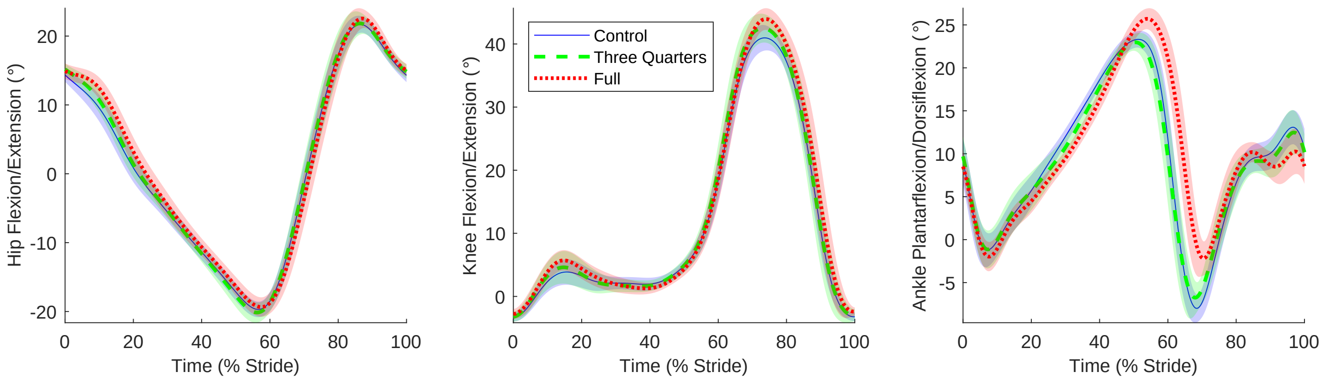

Fig. 5: Sagittal plane hip, knee, and ankle angles from one representative subject from part one. The mean is indicated with the line and one standard deviation is indicated by the shaded region. Hip and knee kinematics are generally unaffected by a foot plate, but ankle ROM is reduced.

in three subjects. The full plate did not significantly affect hindfoot AB/AD on average, although there were a range of individual subject responses. Neither plate had an overall effect on hindfoot inversion (INV/EV) ROM, although there were a range of individual subject responses. The three-quarter plate reduced forefoot PF/DF ROM by 19% (12.4° vs. 15.4° for control) and the full plate reduced it by 21% (to 12.2°). The three-quarter plate did not change forefoot AB/AD for most subjects, but the full plate did decrease forefoot AB/AD ROM by 54% (from 10.2° to 4.7°). Although there was not an overall statistical decrease in forefoot INV/EV for the three-quarter plate when averaged across subjects, three of the four subjects decreased it by 37% (from 11.0° to 7.0°). A full plate decreased forefoot INV/EV ROM by 51% (to 5.4°). Toe PF/DF ROM decreased in two subjects for the three-quarter plate (from 24.4° to 20.2°, a 17% change) and in all four subjects for the full plate (to 13.0°, a 47% change). The effect of a three-quarter plate on toe AB/AD and INV/EV ROM was unclear, with no consistent pattern of responses across subjects. A full plate reduced toe AB/AD ROM by 20% and INV/EV ROM by 29%. Overall, the presence of a foot plate reduced foot segment motion.

One subject in part two had a different response than the other subjects, particularly for the three-quarter plate. In contrast to the other subjects, this subject increased the ROM for almost all foot segments in all three planes with the three-quarter plate. With the full plate, this subject decreased toe PF/DF and AB/AD ROM as well as hindfoot PF/DF, agreeing with the other subjects. However, their changes in the other foot segments differed from the other subjects. This subject had equivalent hip ROM across conditions, as is typical, but increased knee ROM with both foot plates. This subject also walked at $0.8 m/s$ with the foot plates, while the other subjects averaged $1.2 m/s$.

3.3 Ground Reaction Forces

For part one (N=12), vertical GRFs were not significantly affected by either foot plate (Tab. 1, Fig. 7). Similarly, A/P GRF at heel strike was unchanged. The three-quarter plate did not change A/P GRF at push off, but a full plate did significantly decrease it from $2.1 N/kg$ to $1.6 N/kg$ (24% change). In the M/L direction, changes in peak GRF were either equivalent or inconclusive.

Results were similar for part two (N=4) (Tab. 2, Fig. 8). The vertical GRFs for both foot plate conditions were equivalent to the control condition. The A/P GRF at heel strike was equivalent for the three-quarter plate, but inconclusive for the full plate (it increased in two subjects and decreased in two subjects). A/P GRF at toe off was decreased by both plates, from $2.0 N/kg$ to $1.6 N/kg$. Both peak M/L GRFs were reduced with the full plate but were equivalent for the three-quarter plate.

3.4 Energy and Power

The UD model captures the overall power of the foot and foot plate. The UD power curves appeared similar for all conditions in both parts of the study (Fig. 9). In general, the power was low or negative for the first half of the stance period. The UD segment then generated positive power, peaking at around 80% of the stance period before rapidly reducing prior to toe-off. It is unclear if or how energy in the UD segment was affected by the foot plates (Tab. 1, 2). For most subjects, the change in integrated power was statistically inconclusive because of high step-to-step variability.

The three-quarter plate stored very little energy over stance, approximately 0.02 J/kg (Fig. 9c). For comparison, the total energy in the UD segment averaged approximately $1.5 J/kg$ during the first 60% of stance. The full plate also stored very little energy for most of stance, but deflected around push off and stored significantly more energy, up to approximately 0.18 J/kg. This peak in elastic energy occurred at about the same time as the peak in UD power.

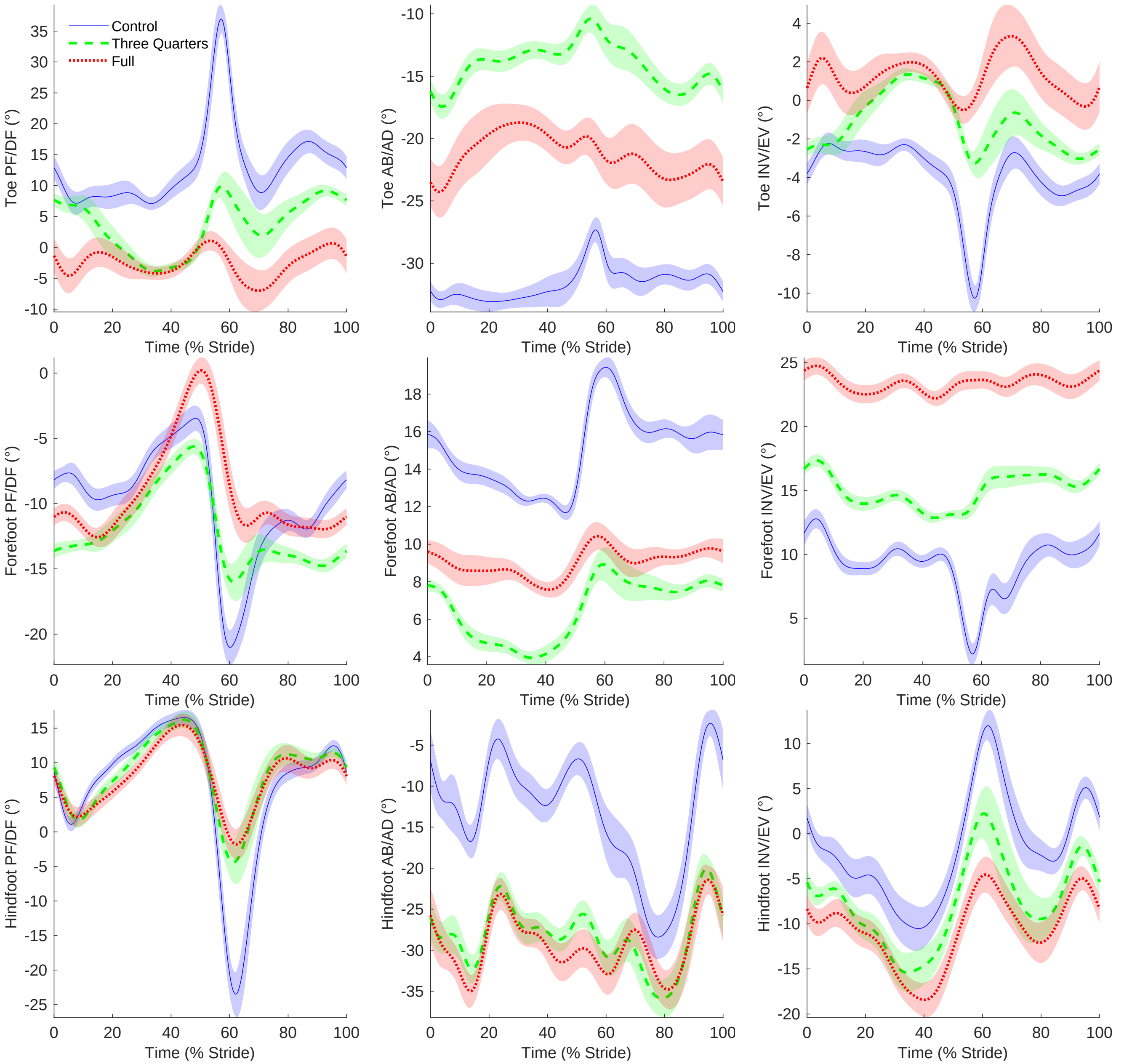

Fig. 6: Intersegmental foot angles of a representative subject from part two. The mean is indicated with the line and one standard deviation is indicated by the shaded region. DF, AB, and INV are positive. As discussed in Sec. 2.2, absolute offsets in the curve between conditions were partly due to markers being re-applied and are not meaningful. Toe off occurs at approximately 60% of stride, which is where the most significant changes between conditions occur. This subject exhibited significantly reduced toe DF and hindfoot PF with a foot plate.

4 Discussion

Neither walking speed nor stride frequency was altered by a foot plate in either part of the study. This is consistent with amputee research in which altering prosthetic foot stiffness did not change walking speed [44]. This suggests that humans adjust foot and ankle behavior in order to compensate for changed foot stiffness to maintain a desired speed. The frequency results are somewhat in contrast to comparisons of barefoot (no extra stiffness) vs. shod gait, because shoes typically change stride frequency [30].

Hip and knee kinematics were generally equivalent across all three conditions, indicating that kinematic compensations occur distal to the knee [25, 45]. While some subjects decreased early stance knee flexion, most did not. This is consistent with AFO studies [32]. A foot plate did consistently decrease ankle ROM when using the PiG model and hindfoot ROM when using the Oxford foot model. This is not surprising since these angles measure approximately the same quantity. Further, the full, effectively stiffer, foot plate reduced ankle ROM more than the three-quarter plate. Many [25, 26, 32] but not

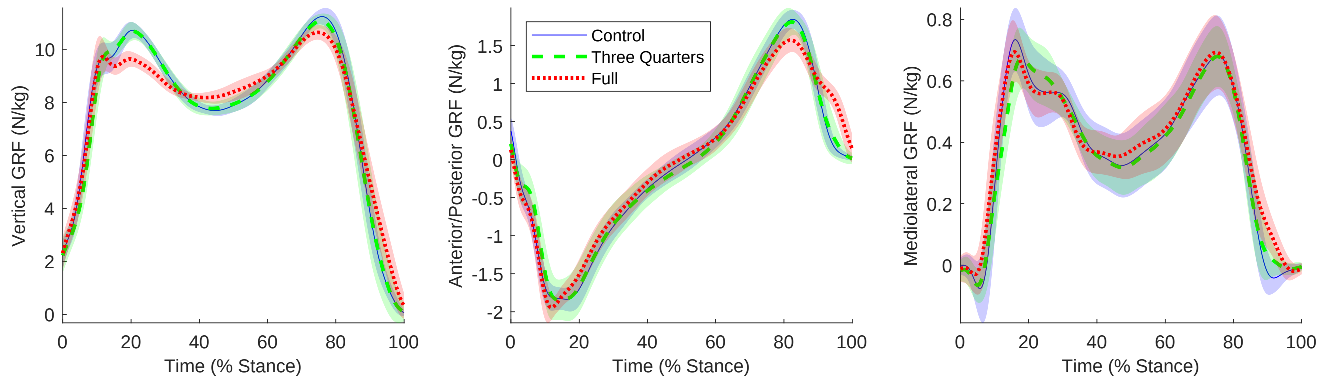

Fig. 7: Ground reaction forces of one representative subject from part one. The mean is indicated with the line and one standard deviation is indicated by the shaded region. With the exception of the peak push-off force in the A/P direction, there were no significant changes in GRF between conditions in part one.

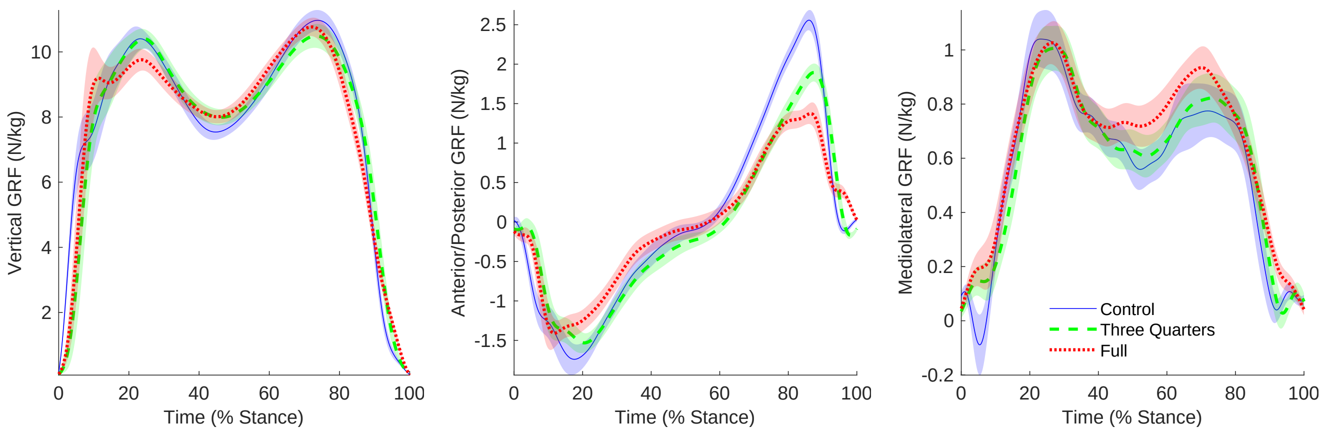

Fig. 8: Ground reaction forces from a representative subject from part two. The mean is indicated with the line and one standard deviation is indicated by the shaded region. The vertical GRF was unchanged, but forward (AP) GRF was significantly reduced by the full foot plate.

all [23] previous studies have also shown that increasing foot stiffness leads to decreased ankle ROM. In contrast to stride frequency, the change in ankle ROM agrees with studies comparing barefoot to shod walking [30].

Overall, the foot deformed less with a foot plate, consistent with other studies on footwear [29]. The full length plate had a greater effect than the three-quarter length plate. The most significant reduction was at the toe. With the three-quarter plate, the toes could bend freely. With the full plate, the toes and the foot plate had to bend together. This likely directly impacted the windlass mechanism. Toe dorsiflexion pulls the PA, which in turn shortens the arch of the foot. The arch is comprised of parts from the forefoot and hindfoot, so less toe dorsiflexion could affect the entire foot. Because the three-quarter foot plate had less of an effect on foot kinematics, this suggests that a shorter foot plate may be preferable for an exoskeleton.

The vertical GRF peaks were equivalent for all three conditions suggesting that humans adjust for changes in foot stiffness to ensure sufficient vertical support. In simulations, increasing prosthetic foot stiffness led to decreased forward propulsion [46]. Supporting this, the full plate decreased forward GRF at push off. During normal walking, the metatarsals and toes distribute the push off force in late stance [47]. Because of the reduced toe flexion, it is possible that the foot plate does not allow the necessary surface area needed to generate the same forward force. Therefore, greatly increasing the rotational stiffness of the toes may require changes in ground reaction forces.

Given that walking speed remains constant, it is not surprising that UD power did not change. This suggests that humans can adjust for foot stiffness and maintain an overall power profile, despite studies showing that changing prosthetic foot stiffness affects power [20]. Most likely, intact human feet dynamically adjust stiffness via muscle activation to compensate for changes, while prosthetic feet have a set stiffness. Compared to subjects in part one (Fig. 9a) and previous overground studies [37, 40, 41], some subjects in part two had significantly higher power (Fig. 9b), possibly caused by the treadmill. This could indicate that treadmill walking causes significant differences in power, or that treadmill data requires different processing.

The fact that a foot plate reduced propulsive GRF

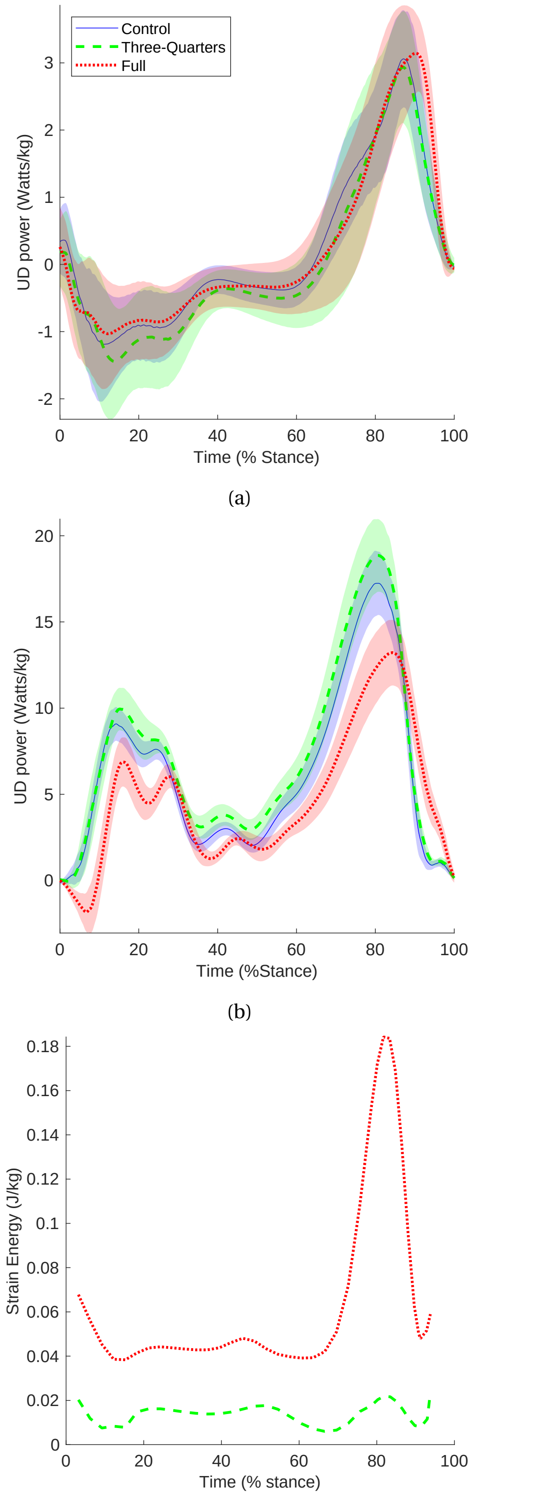

Fig. 9: a) Unified deformable power from one representative subject from part one. b) Unified deformable power from one representative subject from part two. c) FEA results from subject shown in b). For a) and b), the mean is indicated with the line and one standard deviation is indicated by the shaded region. The foot plate stores very little energy for most of stance. Just before toe-off, there is a significant increase in UD power. The toes also flex, causing the full plate to deflect and store much more energy.

and ankle ROM while walking speed remained constant is interesting. Since work is force times distance, these results suggest that the ankle should have performed less work. However, the UD power results indicate that foot and ankle work was unchanged. One possibility is that the foot plate stores and releases the energy that the ankle no longer provides. However, the amount of energy stored in the foot plates appeared to be fairly small, so this seems unlikely. Another possibility is that the ankle was indeed providing a higher force over a shorter distance. Unfortunately, this cannot be verified because it would require an accurate measure of the force distribution between the foot and foot plate.

5 Conclusion

A foot plate in and of itself had surprisingly little effect on gait spatiotemporal parameters or on hip and knee joint kinematics. This rejects part of our main hypothesis. However, the foot plate decreased ankle ROM and foot intersegmental motion, which supports the other part of our main hypothesis. The full plate had a greater effect, which supports our secondary hypothesis. The full plate also reduced the forward push off force. To minimize gait alterations, foot plates should allow the toe to function naturally. If a full length plate is used, the effects will likely be limited to changes in foot and ankle motion.

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7 Conflicts of Interest

The authors do not have any conflicts of interest.

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