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Ankle Kinematics Describing Gait Agility: Considerations in the Design of an Agile Ankle-Foot Prosthesis

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Abstract— **The designs of available lower extremity powered prostheses are focused on a single degree of freedom (DOF) in sagittal plane, allowing the control of their ankle joints in dorsiflexion and plantarflexion. The human gait however, shows that the ankle movements in both sagittal and frontal planes are significant even during walking on a straight path. Additionally, there is a significant change in the ankle movements during straight walking compared to turning and cutting, especially in frontal plane. A better understanding of the ankle characteristics in both sagittal and frontal planes may result in the design of significantly more effective lower extremity prostheses that mimic the ankle function and improve the agility of gait.**

In this paper, the ankle rotations are estimated during step turn and cutting to provide evidence for necessity of a multi-axis design while providing the preliminary design parameters for a prototype multi-axis powered ankle-foot prosthesis. It is shown that the proposed cable-driven prototype is capable of closely mimicking the ankle movements in both sagittal and frontal planes during turning and walking on a straight path.

I. Introduction

Many gait scenarios such as traversing slopes or turning requires agile movements of the ankle in both sagittal and frontal planes. Agility is defined as the ability to move quickly and easily [1] and it is fundamental for a natural and efficient gait. Agility is essential when changing directions or accommodating disturbances on the terrain to minimize energy consumption and reduce the risk of injury.

Recent advances in powered prostheses promise to significantly improve the quality of life of individuals with impaired mobility. A better understanding of the complexities surrounding lower limb prostheses, will lead to increased health and well-being for the 1.7 million limb amputees in the US, the majority of whom have lower extremity amputations [2, 3]. Currently commercially available powered ankle-foot prostheses are capable of controlling a single DOF in the sagittal plane, focusing on improved mobility in straight walking even though turning steps represents an average of ~25% of steps taken during a typical day [4]. Because current prostheses are not designed to assist turning, amputees and non-amputees exhibit different turning strategies. During turning non-amputees typically generate most propulsion at the ankle and the hip movement in the coronal plane. In contrast, amputees using a passive prosthesis generate propulsion by moving the hip in the sagittal plane. It is suggested that such difference in gait strategies are due to lack of sufficient power in the prosthetic ankle and the amputees' desire to prevent fall [5-8]. Such differences in gait strategies lead to a different biomechanics of turn and increased risk of secondary complications. During a turn, ground reaction forces are modulated to accelerate the center of mass of the body along the path; thus, during a step turn, lateral and propulsive impulses are larger compared to a straight step [9]; also, preliminary studies have shown an increase in inversion during a step turn, leaning the body toward the inside of the turn, when compared to a straight step [10]. These evidences suggest that turning may not be considered a passive mechanism and requires modulation of ankle impedance in both sagittal and frontal planes. Therefore, we theorize that an ankle-foot prosthetic robot capable of generating torques in both the dorsiflexion-plantarflexion (DP) and inversion-eversion (IE) directions with impedance modulation similar to the human ankle may improve the user's agility and increase mobility while reducing the risk of secondary injuries or falls.

Understanding of the ankle's capability in impedance modulation and generating net positive work during the stance period of gait has influenced the design of new ankle-foot prostheses [11-14]. One design approach is based on storing energy during the heel strike and releasing it during the push-off before the trailing foot's heel strike. Collins and Kuo [15] developed a microprocessor-controlled artificial foot that limits the increase in metabolic cost to 14% compared to 23% that occurs with a passive prosthesis. On the other hand, there are powered prostheses capable of injecting energy to the system. Sup et al. developed a powered transfemoral prosthesis with active knee and ankle joints, each with one controllable DOF in the sagittal plane [16-19]. The controller adjusts the impedance at a number of instants during gait by altering the neutral position of the ankle. Au et al. developed the ankle-foot prosthesis BiOM [20-22], which provides the necessary energy during push off and generates a net positive work [23, 24] that has been shown to reduce the metabolic costs by 8.9% to 12.1% at different gait speeds compared to a passive prosthesis and increased the preferred gait speed by 23% [25].

While the aforementioned prostheses have advanced the state-of-the-art, their designs are confined to the sagittal plane. Even level walking in a straight line requires the ankle to function in both the sagittal and frontal planes. Additionally, normal daily activity includes more gait scenarios which requires agile movements such as turning, traversing slopes, steering, and adapting to uneven terrain profiles. This suggests

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that the next advancement in lower extremity assistive devices is to extend their design and control to the frontal plane. To do that, we need a better understanding of the multi-variable mechanical impedance of the human ankle which requires knowledge about the time history of the ankle angles and torques during different gait scenarios. The ankle displacements needs to be studied since the mechanical impedance of the ankle is a dynamic operator that maps the time-history of angular displacements onto the corresponding time-history of torques at the ankle joint. In this paper, the ankle angles during straight walk and different turning scenarios were measured. We use the term step turn to describe the maneuver used to change the walking direction by pivoting around the leading leg and rotating into a new direction (approximately perpendicular to the initial direction such as turning around a corner). Sidestep cutting is used to describe the motion of pushing the body sideways using the leading leg to translate the body while walking forward (the motion is at near 45 degrees from the original path) without rotating the body (e.g. stepping sideways to avoiding an obstacle on the ground). We introduced the concept of a multi-axis powered ankle-foot prosthesis and showed the capability of this concept to mimicking the ankle angles. We first described the experiments for collecting the information on the ankle angles during different gaits. Next, the design and testing of the proof of concept prototype of an ankle-foot prosthesis with two DOFs will be explained.

II. Ankle Rotations During Gait

To change the direction in gait, one needs to perform different gait maneuvers such as step turn, spin turn, or sidestep cutting that have different kinematics. For example, compared to straight walking, step turns have considerably different velocity, length, width, and higher turning reaction forces [2, 6, 8, 9]. Also the ankle moment in the inversion direction is significantly different from the straight steps and spin turn steps[26].

A series of experiments were performed to quantify the kinematic behavior of the ankle in the context of agility of gait. The experiments measured the ankle rotations during stance period of step turn and sidestep cutting and compared the results to the ankle rotation during straight walking. The study however, did not include any cognitive aspect of the agility, but focused on the kinematics of the gait due to change of direction and speed. Additionally, the ankle rotations were used to provide design parameters for the range of motion (ROM) of the prosthesis and to evaluate the kinematic design of the ankle-foot prosthesis in reproducing the same trajectories.

There have been different approaches to measure ankle rotations during gait that include using flexible electro-goniometer, electromagnetic tracking devices, and motion capturing cameras [6, 8, 9, 26, 27]. We used a motion capture camera system to track the three-dimensional rotations of the foot and tibia in stance periods. The motion capture camera system consisted of eight cameras in a square formation covering a volume of about 16 cubic meters and an area of 12 square meters. The cameras emitted infrared light and captured the reflected light from reflectors mounted on the participants with a rate of 250 Hz. Reflective markers were attached to polycarbonate plastic rigid bodies. One rigid body was attached to the participant's shin resting against the tibia bone to record the shin rotations. Another rigid body was attached to the user's shoe above the metatarsal bones to record the foot rotations. The ankle rotations were calculated as the relative angles between the foot and shin.

Subjects with no self-reported neuromuscular and biomechanical disorders were recruited for the experiments. The subjects gave written consent to participate in the experiment that was approved by the Michigan Technological University Institutional Review Board. The experiments with two sets of gait scenarios were performed: 1- Step turn, and 2- Sidestep cutting. The details of the experiments and the results follow.

A. Ankle Rotations during the Stance Period of Step Turn

Straight walking requires a complex sequence of muscle activation to modulate the ground reaction forces to produce forward motion. Similarly, modulation of the reaction forces is required for turning the body [5]. Two different strategies that are commonly used for turning are the spin turn and the step turn. The spin turn consists of turning the body around the leading leg (e.g. turning right with the right leg in front). The step turn consists of shifting the body weight to the leading leg and stepping onto the opposite leg while still shifting the body weight (e.g. turning left with the right leg in front). It has been shown that the step turn velocity, length, and width are considerably different than the straight walk with higher turning reaction forces [9]. Three-dimensional measurement of the ankle angles during step and spin turns have been previously studied [26]; however, it is of interest to study the ankle angular displacements during different phases of the stance period of turning steps and compare these results to the ankle angles during straight steps.

Five male subjects participated in this study. The subjects were instructed to walk at a normal pace with an audible metronome synchronized to their number of steps per minute in an attempt to keep the walking speed constant. The gait speed for the participants ranged from 88 to 96 steps per minute. They started walking from outside the field of view of the cameras while following a straight line marked on the floor (Fig 1). When they reached a reference point on the floor, they performed a 90° step turn to the left, pivoting on their right leg and continued walking straight until they were outside the field of view of the cameras. Each subject repeated the test nine times, after several training trials to increase the consistency of the trials. Time trajectories of the markers on the tibia and foot were used to estimate the ankle rotations in DP, IE and medial-lateral (ML) directions. The data for each test was divided into 6 phases: Heel strike (consists of heel strike and loading response), mid stance, and push off (consisted of terminal stance and pre-swing phases) for both straight and turning steps. The averages of the DP, IE, and ML rotations of each phase were calculated for all 9 trials of 5 subjects (a total of 45 trials).

Table 1 shows the average ROM of the subjects during the stance periods of straight step and step turn. Table 2 shows the average rotations and the difference in angles from the turning step to the straight step in each phase. The ROM of each subject's ankle about the three axes of the ankle and their average rotations during the stance periods were calculated and used to find the average percent change from straight walk to step turn with respect to their ROM during the straight step (Table 2).

Table 1 shows a modest decrease of ROM in DP direction during the step turn compared to the straight step. The ROM in the IE direction increased by 23.8%, indicating the significance of the IE role during turning. A significantly smaller ROM in ML may suggest a higher stiffness in that axis of rotation necessary to transfer the reaction forces from the ground to the body. As the step progressed through the gait cycle, noticeable differences were observed between the straight step and step turn for all subjects.

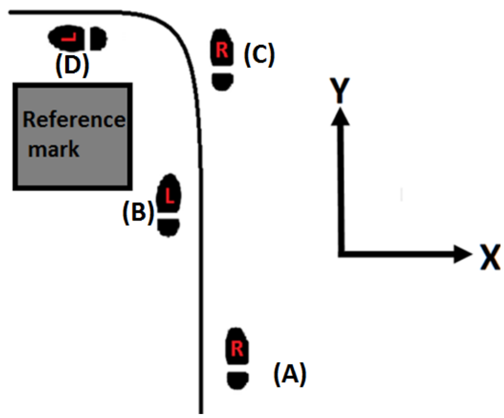

Figure 1. Foot positions during step turn test. (A) straight step right ankle. (B) straight step left ankle. (C) turn step right ankle. (D) post-turn left ankle.

TABLE I. ANKLE ROM (IN DEGREES) THROUGHOUT STANCE FOR STRAIGHT WALK AND STEP TURN

	Straight Step mean (std. error)	*Step turn mean (std. error)*	*Angular Change*	*Percent Change*
DP	33.9 (0.7)	31.6 (0.6)	-2.3	-7.4
IE	15.7 (0.5)	20.6 (1.1)	4.9	23.8
ML	22.1 (0.6)	16.8 (0.7)	-5.3	-31.9

TABLE II. AVERAGE ANKLE ROTATIONS (IN DEGREES) DURING STANCE PHASES OF STRAIGHT STEPS AND STEP TURNS

		Straight Step Mean (std. error)	*Step turn Mean (std. error)*	*Angular Change*	*Percent Change as a Percent of ROM**
DP	***heel strike***	-8.7 (0.8)	-9.7 (1.0)	-1.0	-2.8
	mid stance	2.3 (0.6)	0.4 (0.6)	-2.0	-5.8
	push off	10.6 (1.2)	1.4 (0.9)	-9.2	-27.2
IE	***heel strike***	-1.7 (0.5)	5.9 (0.6)	7.6	48.5
	mid stance	-2.9 (0.3)	6.5 (0.2)	9.4	60.1
	push off	1.4 (0.5)	13.6 (0.5)	12.2	77.5
ML	***heel strike***	-5.3 (0.6)	0.3 (0.6)	5.7	25.7
	mid stance	-0.9 (0.5)	-3.6 (0.4)	-2.7	-12.0
	push off	5.5 (0.3)	-6.5 (0.7)	-12.1	-54.6

* Angular change as a percent of the corresponding average ROM of straight step.

Table 2 shows the average ankle rotations in straight and turning steps at different phases of stance periods. During a step turn, IE had the largest deviation from the ankle rotations in the straight step. During the step turn, IE started with 5.9° of inversion and increased to 13.6° at push off, suggesting a gradual increase in inversion to lean the body toward the inside of the turn. This was significantly different from straight step that started at 1.7° eversion at heel strike and transitioned to 1.4° inversion at push off. These results indicated that the change in ankle angle in the IE direction at the step turn is significantly larger and different from straight step [10]. The ankle inversion is required for generating a ground reaction force during the step turn as reported in [9, 28]. DP displacement started at a similar initial angle as the straight step at the heel strike (-9.7° of dorsiflexion) but progressively showed less plantarflexion at push off (1.4° in step turn compared to 10.4° in straight walk). At the heel strike of the step turn, ML displacement had an increase of 5.7° of medial rotation compared to straight walk that may suggest an anticipatory motion of the foot. The difference in lateral rotation during straight step and step turn at the push off increased to 12.1°.

B. *Ankle Rotations of Sidestep Cutting and Step Turn at Different Gait Speeds*

In a second set of experiments, step turn and sidestep cutting maneuvers at two different gait speeds were studied and the results were compared to the ankle rotations in straight steps in both left and right ankles. Seven young subjects were participated in this set of experiment. The slow speed was set to 96 steps per minute synchronized to an audible metronome. The fast speed was different among the participants with an average of 114 steps per minute, calculated from the right foot data. The subjects were instructed to go as fast as they felt comfortable to perform the step turn and sidestep cutting without occurrence of a flight phase [29]. The step turn experiments at both speeds were performed with similar protocol to the previous experiment. For the sidestep cutting experiments, the participants were instructed to walk straight from the outside the field of view of the cameras. When they reached a set of obstacles on the ground, they performed a sidestep cutting to the left, pivoting about their right leg to avoid the obstacle and switching direction immediately; followed by a left sidestep cutting that redirects the walk in a straight line parallel to the initial direction of gait (Fig. 2). Each experiment was repeated five times for each subject and the results were averaged across the trials. The average angular rotations of each stride segment were calculated across the participants' data. Both experiments showed that the percent change in IE direction were greater than the other two DOFs, confirming the results from the previous experiment.

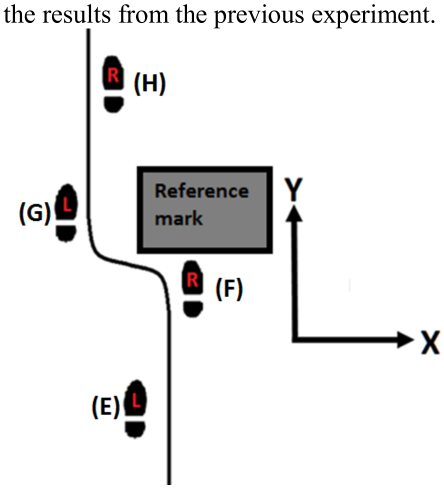

Figure 2. Foot position during sidestep cutting. (E) pre-cutting left ankle. (F) sidestep cutting right ankle. (G) sidestep cutting left ankle. (H) post-cutting right ankle.

Table 3 shows the average percent change of the ankle rotations in the stance period of right and left ankles during a step turn compared to straight steps at different gait speeds. The step turn initiated on the right foot to redirect the body to the left. It was seen that the IE motion of the right ankle for the straight steps in both slow and fast speed at the corresponding phases are close. However, the deviation of the right ankle IE during the turn was increased significantly with the speed. Specifically, the push off at straight step and low speed showed a 2.8° eversion, while during the turn, it changed to 10.5° inversion. During the fast speed, these values were 2.5° eversion and 14.3° inversion, respectively. Those values were equivalent of 248% and 312% deviations (as a percent of the straight step IE average ROM) from the straight step rotation in IE for the slow and fast speed tests, respectively. Similar trends were observed for heel strike and flat foot of the right ankle.

The behavior of the right ankle during sidestep cutting was similar to the step turn, but the deviation from the straight step is less pronounced compared to the step turn. For the cutting maneuver, the subjects turned slightly left to avoid an obstacle and continued in the initial walking direction. The results are shown in Table 4. A significantly different behavior is seen in the left leg sidestep cutting on the left ankle when the ankle receives the body weight after its redirection. The most significant deviation occurred at the push off of the left ankle, when the weight of the body transferred to the left leg and redirecting the body to the path parallel to the initial direction were initiated. The push off at straight step and low speed showed a -6.9° eversion, while at the step after redirection, it changed to 1.4° inversion. During the fast speed, these values were -8.9° eversion and

TABLE III. BILATERAL ANKLE ROM IN IE DIRECTION DURING STANCE FOR A STEP TURN COMPARED TO STRAIGHT STEPS AT SLOW AND FAST SPEEDS.

	Right Ankle Straight Step, Slow mean (std. error)	***Right Ankle Turning Step, Slow mean (std. error)***	***Angular Change as a Percent of ROM****
heel strike	-2.3 (0.8)	7.8 (0.9)	180
mid stance	-3.4 (0.7)	7.3 (0.8)	199
push off	-2.8 (0.7)	10.6 (1.3)	248
	Right Ankle Straight Step, Fast	***Right Ankle Turning Step, Fast***	
heel strike	-2.5 (0.9)	9.9 (0.9)	231
mid stance	-3.1 (0.7)	10.8 (0.7)	259
push off	-2.5 (0.8)	14.3 (1.0)	312
	Left Ankle Straight Step, Slow	***Left Ankle Turning Step, Slow***	
heel strike	-4.8(1.2)	-7.2 (2.0)	-45
mid stance	-7.3 (1.3)	-8.0 (1.4)	-13
push off	-9.7 (1.6)	-8.5 (1.4)	22
	Left Ankle Straight Step, Fast	***Left Ankle Turning Step, Fast***	
heel strike	-5.0 (1.2)	-8.5 (1.6)	-66
mid stance	-7.8 (1.5)	-9.1 (1.5)	-24
push off	-9.5 (1.8)	-8.5 (1.3)	20

* Angular change as a percent of the corresponding average ROM of straight step

TABLE IV. BILATERAL ANKLE ROTATIONS IN IE DIRECTION DURING STANCE PHASES OF SIDESTEP CUTTING COMPARED TO STRAIGHT STEPS AT SLOW AND FAST SPEEDS.

	Right Ankle Straight Step, Slow mean (std. error)	***Right Ankle Cutting Step, Slow mean (std. error)***	***Angular Change as a Percent of ROM****
heel strike	-2.3 (0.8)	-1.8 (1.1)	9
mid stance	-3.4 (0.7)	-2.1 (0.9)	25
push off	-2.8 (0.7)	0.8 (1.1)	68
	Right Ankle Straight Step, Fast	***Right Ankle Cutting Step, Fast***	
heel strike	-2.5 (0.9)	-0.2 (1.0)	43
mid stance	-3.1 (0.7)	-0.2 (1.0)	55
push off	-2.5 (0.9)	1.9 (1.4)	81
	Left Ankle Straight Step, Slow	***Left Ankle Cutting Step, Slow***	
heel strike	-1.9 (1.0)	5.3 (1.1)	134
mid stance	-4.7 (0.9)	1.6 (0.9)	118
push off	-6.9 (1.1)	1.4 (1.1)	154
	Left Ankle Straight Step, Fast	***Left Ankle Cutting Step, Fast***	
heel strike	-4.4 (1.6)	5.0 (1.5)	174
mid stance	-7.4 (1.7)	2.9 (1.4)	191
push off	-8.9 (2.2)	2.7 (1.6)	215

* Angular change as a percent of the corresponding average ROM of straight step

2.7° inversion, respectively. Those were equivalent of 154% and 215% deviations of IE at the heel strike for the slow and fast speed, respectively.

III. MULTI-AXIS ANKLE-FOOT PROTOTYPE

The result from the tests of the ankle rotations in three DOFs suggested that a multi-axis mechanism in a prosthesis may enhance gait efficiency by extending the control of IE during turning and cutting. This novel design is anticipated to enable the device to adapt to uneven and inclined ground surface conditions and allow the amputees to benefit more from their prostheses rather than using their hip joint; enabling a more agile and natural gait with less stress on other joints. A prototype design of a cable-driven ankle-foot prosthesis with two controllable DOFs was designed and fabricated in an effort to study the feasibility of the steering and maneuverability requirements from a 2 DOFs ankle (Fig. 3). The design aimed to allow the same ROM and angular velocity in straight walking and turning as the human ankle while producing enough torque for propulsion.

The device consists of two DC motors (A) and planetary gear heads (B) powered by two motor controllers (C) connected to two quadrature encoders (D). Two cable drums (E) transfer the required torque to the ankle through the shock-absorbing nylon rope (F). A universal joint (G) connects the pylon to the foot and an elastic carbon-fiber plate. Both actuators apply the torque to the foot using a cable-driven mechanism with pulleys (H). The cable is attached to a carbon fiber plate (I) which is connected to a commercially available prosthetic foot (Otto Bock Axtion®) (J). In the rear side of the carbon fiber plate, the cable is mounted at both sides of the longitudinal axis of the foot. At the front side of the carbon fiber plate, the cable is passed through a pulley (K). The mechanism is capable of both DP when the motors rotate in opposite directions and IE when the motors rotate in the same direction. Also, any combination of DP and IE can be obtainable by combining different amounts of rotation in each motor.

Currently, two optical quadrature encoders (200 pulses per revolution) provide position feedback to a remote computer that uses a proportional plus rate controller to control the relative position of the foot with respect to the pylon. To evaluate the capability of the mechanism to reproduce ankle rotations similar to the human ankle during the gait, look-up tables with recorded data of a representative subject were used by the controller. For the stance and swing periods of the right ankle during the step turn at fast speed, the time trajectories of output angles and the data recorded from the human ankle rotations in DP and IE directions are shown in Fig. 4. Similarly, the time trajectories of the right ankle during the sidestep cutting at fast speed are shown in Fig. 5 and the time trajectories of the left ankle during the sidestep cutting at fast speed are shown in Fig. 6. All signals were filtered with a low-pass filter with a cutoff frequency of 5 Hz to remove sensor noise from the output signal. During the tests, the robot was moving at 50% of the fast human walking speed (57 steps per minute). For ease of comparison, the output plots have a time shift to remove a delay of 56 milliseconds (Fig 4 and 5) and 24 milliseconds (Fig. 6). The right ankle during the step turn and the left ankle during the sidestep cutting were subject to larger angular displacements when compared to the right ankle in the sidestep cutting, resulting in the observed increased delay in the prosthesis. The current prototype was developed as a proof of concept to validate the design kinematics; therefore, faster motors and sensors with lower noise levels will be used in future designs. All plots indicate that the mechanism is capable of reproducing the same ankle rotations as the human subjects during step turn and sidestep cutting.

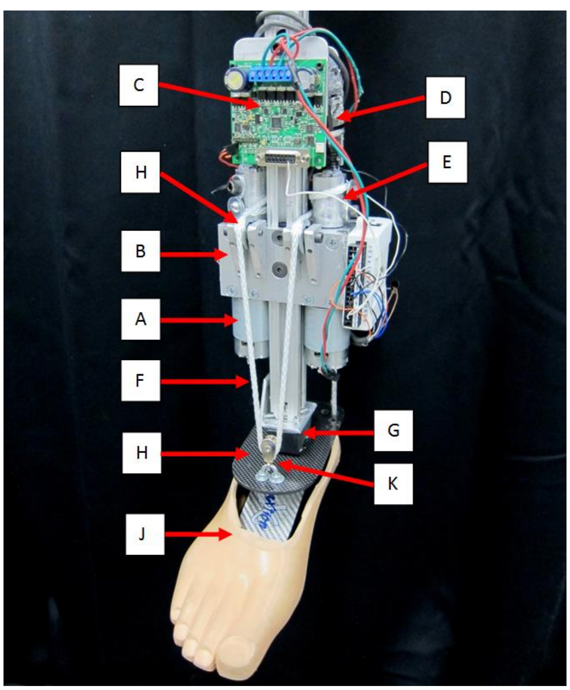

Figure 3. Prototype of the powered ankle-foot prosthesis with two DOFs.

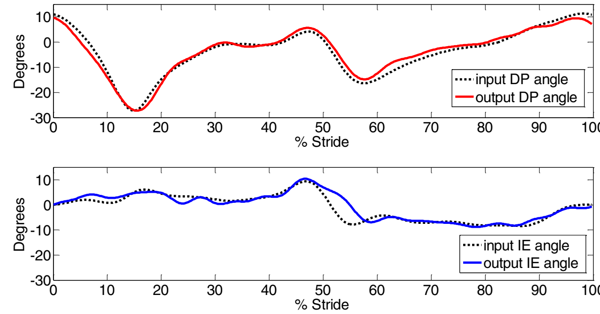

Figure 4. Input and compensated output for time delay (56 milliseconds) of the ankle-foot prosthesis. The input is the recorded right ankle rotations of a representative subject during swing and stance periods of the step turn at fast speed.

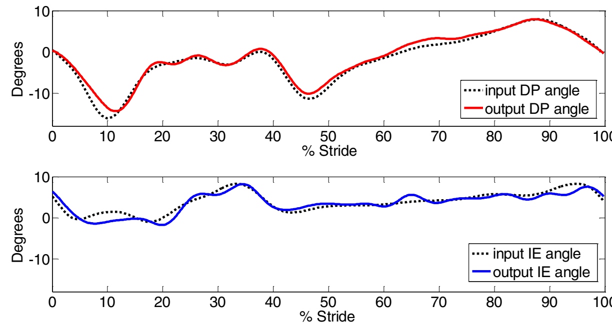

Figure 5. Input and compensated output for time delay (24 milliseconds) of the ankle-foot prosthesis. The input is the recorded right ankle rotations of a representative subject during swing and stance periods of the sidestep cutting at fast speed.

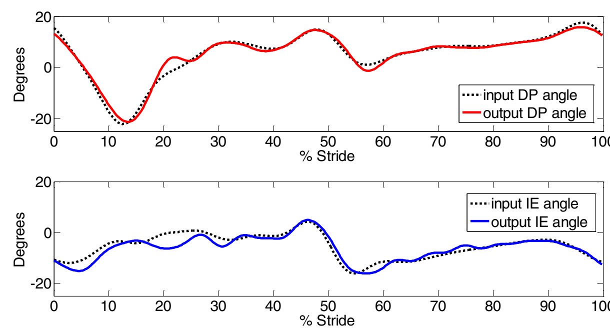

Figure 6. Input and compensated output for time delay (56 milliseconds) of the ankle-foot prosthesis. The input is the recorded left ankle rotations of a representative subject during swing and stance periods of the sidestep cutting at fast speed.

IV. Conclusion

Human ankle rotation during step turn and sidestep cutting at two different speeds were measured using a camera system. It was shown that the rotation of the ankle in inversion-eversion significantly changed during those gait maneuvers when compared to straight walking. The results implied that a multi-axis ankle-foot prosthesis could increase the agility of the gait by mimicking the ankle kinematics. The results were used as design parameters for fabrication of a prototype cable-driven powered ankle-foot prosthesis with two degrees of freedom. Evaluation experiments showed that the mechanism is capable of reproducing the human ankle rotations during step turn and side step cutting, suggesting the feasibility of the design in mimicking the dynamics of the human ankle.

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