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Control of a 2-DOF powered ankle-foot mechanism

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***Abstract*— This paper describes a finite state machine to control an ankle-foot prosthesis with two degrees of freedom (DOF) in the sagittal and frontal planes. Strain gauges were installed in the foot to provide ground reaction torques feedback for impedance and admittance controllers to be used at heel-strike and push-off of the gait, respectively. The quasi-static stiffness of the ankle with the active control was measured showing a near linear relationship between the torque feedback gain and the stiffness of the ankle. The performance of the finite state machine and controllers were also evaluated using a custom-made circular treadmill and the results were compared to the results of the prosthesis using position controller and inactive controllers. The results showed that the impedance/admittance controller was capable of tracking the desired input trajectory while decreasing the required torque at the ankle joint.**

I. INTRODUCTION

The development of powered prostheses has been focused on forward motion; however, daily activities consist of an average of 25% turning steps [1]. Turning requires control of the ankle function in both the frontal and sagittal planes to control the lateral and forward reaction forces to maintain the body's center of mass along the desired trajectory. This results in increased lateral and forward forces when compared to the straight walking [2].

When physical systems interact with each other, they behave either as an impedance (e.g. accepts external motion inputs and generates force outputs) or an admittance (e.g. accepts external force inputs and generates motion outputs) [3]. Two coupled mechanical systems physically complement each other, when one system is an admittance component the other is an impedance component [3]. During gait, at heel-strike the ankle is manipulated by the environment accepting the external force and generating motion, thus the ankle may be considered an admittance system to match the environment's impedance. At push-off, the ankle may be considered as an impedance system that generates the required torque to produce the desired motion in the admitting environment. This suggests that ankle-foot prostheses could benefit from an admittance controller at heel-strike and an impedance controller at push-off.

An impedance controller uses position encoders to estimate the position of the system. Based on the position, the controller estimates the required torque to generate the desired motion. Torque sensors are used to provide external torque feedback that along with the desired torque, are used to define the appropriate input to the motors [4]. An admittance controller, also called position based impedance controller, uses the environment torque feedback to estimate the appropriate actuator position. The desired actuators' position and position feedback are used to estimate the appropriate actuators inputs [4].

The quasi-static stiffness of the ankle (the ankle's stiffness at relatively slow angular speed) has been estimated [5] and used in the design of ankle-foot prostheses capable of producing net positive work in the sagittal plane. Sup et al. developed a powered knee and ankle prosthesis capable of controlling the mechanical impedance of both the knee and ankle in the sagittal plane [6, 7]. BiOM is a commercially available ankle prosthesis capable of providing the necessary energy during push-off (plantarflexion) [8]. The controllers in both of the aforementioned prostheses use a finite state machine to identify the gait phase to estimate the appropriate ankle torques. This allows the prostheses to adapt to different gait scenarios such as different cadence or slopes. Although the aforementioned prostheses improve the gait characteristics, they are designed to generate the ankle torques in the sagittal plane only.

In this paper the concept prototype of a powered ankle-foot prosthesis capable of controlling two degrees of freedom is presented. The paper describes the use of strain gauges to estimate ground reaction torques and the development of admittance and impedance controllers for the ankle-foot prosthesis based on position and torque feedbacks. The controller adjusts the neutral position of the foot based on pre recorded kinematics of the human ankle during normal walk and switches between admittance and impedance controllers at heel-strike and push-off, respectively using a finite state machine. Also, preliminary tests to evaluate the impedance controller and the performance of the prosthesis while walking on a circular treadmill as an evaluation platform are presented.

II. CABLE-DRIVEN POWERED ANKLE-FOOT PROSTHESIS WITH TWO CONTROLLABLE DEGREES OF FREEDOM

A prototype cable-driven ankle-foot prosthesis with controllable frontal and sagittal planes functions was developed in an effort to study steering and maneuvering with a 2-DOF ankle (Fig. 1). The design allowed for the same range of motion and angular velocity as the human ankle during straight walking and turning and it produces enough torque for propulsion. The mechanism is capable of moving in Dorsiflexion-Plantarflexion (DP) when the motors rotate in

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opposite directions and in Inversion-eversion (IE) when the motors rotate in the same direction. Also, any combination of DP and IE can be obtained by combining different amounts of rotations in each motor.

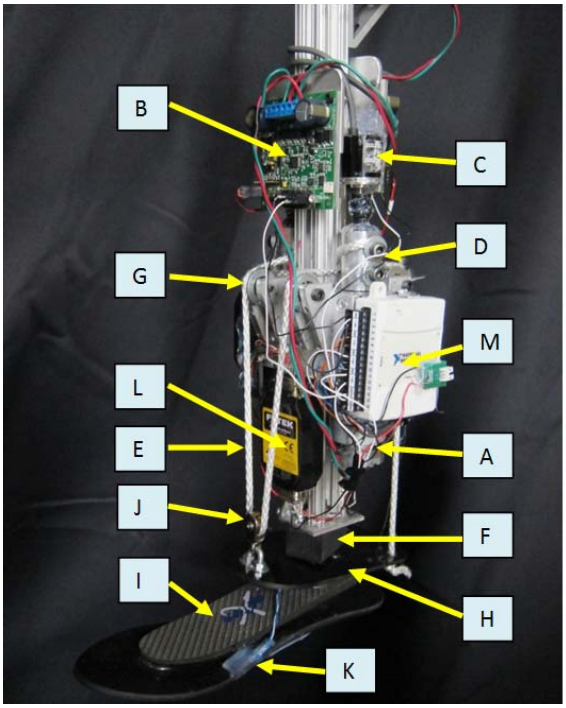

Figure 1. Two-DOF ankle-foot prosthesis prototype. The device consists of two DC motors and planetary gear heads (A) powered by two motor controllers (B) connected to two quadrature encoders (C). Two cable drums (D) transfer the required torque to the ankle through the shock-absorbing nylon rope (E). A universal joint (F) connects the pylon to the foot and is surrounded by an elastomer to provide passive stiffness and damping to the ankle. Both actuators apply the torque to the foot using a cable-driven mechanism with pulleys (G). The cable is attached to a carbon fiber plate (H), which is connected to a commercially available prosthetic foot (Otto Bock Axtion®) (I). In the rear side of the carbon fiber plate, the cable is mounted to both sides of the longitudinal axis of the foot. At the front side of the carbon fiber plate, the cable passes through a pulley (J). Torque feedback is provided by six strain gauges in the foot (K) using two strain gauge amplifiers (L). An analog to digital converter (M) is connected to a remote computer and is used to acquire the sensors data and provide the motor controllers' inputs.

III. Ankle Torque and Angle Feedback

To develop impedance and admittance controllers both force and position feedbacks are required. The angular displacement of the ankle in DP and IE (θ_{DP} and θ_{IE}, respectively) can be calculated from the left and right quadrature encoders' feedback (θ_{Left} and θ_{Right}, respectively), where K_{DP} and K_{IE} are constant gains to define the ankle rotations as a function of the quadrature encoders' outputs, and are based on the prosthesis geometry.

$$\theta_{DP} = K_{dp}\left(\frac{\theta_{Left} + \theta_{Right}}{2}\right) \tag{1}$$

$$\theta_{IE} = K_{ie}\left(\frac{\theta_{Right} - \theta_{Left}}{2}\right) \tag{2}$$

The estimation of the ground reaction torques hinges on the two Wheatstone Bridges (WB) mounted on the foot as described in [9] . Briefly, two Wheatstone Bridges (WB) were used; one for DP and one for IE. In DP, four strain gauges were attached to the sole of the foot. Of those, two strain gauges were located behind the center of rotation of the ankle in the sagittal plane and were wired into opposite sides of the DP WB. The other two strain gauges were located in front of the center of rotation of the ankle in the sagittal plane and were wired to the remainder sides of the WB. Any ground reaction force at the heel caused a decrease in voltage of the WB, which could be correlated to the torque in DP at heel-strike. Any ground reaction force from the ground at the front of the foot caused an increase in the output voltage of the WB, which could be correlated to the torque in DP of the foot during push-off. Note that when the foot was flat on the ground the output from the strain gauges in the heel cancel the output from the strain gauges in the forefoot; therefore, the resultant voltage could always be correlated to the net DP torque in the ankle. In IE, also four strain gauges were used in a WB. Of those, two strain gauges were attached to the top of the forefoot and wired to IE WB. The other two strain gauges of this WB were attached to an inert carbon fiber plate. The strain gauges were placed on the outside edges of the foot and were wired to the same side of the WB (both wired to the same end of the excitation voltage, and connected to the opposite ends of the voltage collector); hence, the difference in strains caused an increase or decrease in voltage at the WB. The voltage can be correlated to the IE torque in the foot when the front of the foot is contacting the ground (e.g. push-off). This configuration made the IE torque measurement decoupled from the torque in DP since if both strain gauges contract or stretch by the same amount, as it would happen in the presence of a DP torque, the output from the WB would not be affected.

To calibrate the output of the WB to the external toques a Kistler® Type 5233A force plate was used to measure external forces applied to the foot. Based on the foot geometry, the point where the force is applied to the foot, the magnitude of the measured forces, and the observed output voltages from the WB, the calibration factors were calculated. For DP, the calibration factor at heel loading and forefoot loading were 1.41 Nm/volt and 19.52 Nm/volt, respectively. The difference was expected as the strain gauges were attached to two different areas of the prosthetic foot. For IE, the calibration factors for inversion and eversion torques were 4.43 Nm/volt and 3.55 Nm/volt, respectively. Close factors were expected since the foot is near symmetrical in the sagittal plane.

IV. Controller

A. Finite State Machine

A finite-state machine was developed to use pre-recorded time-history of the ankle angles in the frontal and sagittal planes during normal walk and real-time torque feedback from the WB to switch between impedance and admittance controllers. The recorded ankle angles of an unimpaired human subject were measured using a motion capture camera system and were accessible as a look-up data table to the estate machine and controllers.

Figure 2 shows that the foot starts at the middle of the swing phase and moves with the impedance controller to the expected heel-strike. If heel-strike is detected before it is expected (e.g. the user started to walk faster) the finite state machine skips the rest of the swing phase and starts the heel-strike phase immediately with the admittance control. If it does not detect a heel-strike (e.g. the user started to walk slower or has stopped), the prosthesis will advance to the position at the beginning of the heel-strike phase and will hold that position until a heel-strike is detected. This is important as the system can adjust to small variations in walking speed, and can start and stop automatically when the gait cycle begins or ends. Once the foot reaches foot-flat phase, the control switches back to the impedance controller until the foot reaches the middle of the swing phase where the cycle (index I in Fig. 2) is reset to zero and it starts again. Note that in admittance control the increment of the index (I) is a function of the external torque disturbance T_{DP}, which will be explored later.

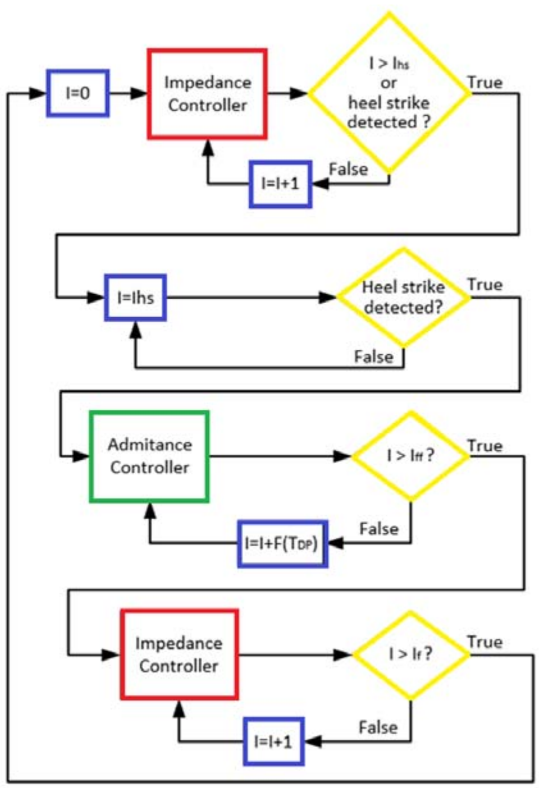

Figure 2. Finite-state machine to switch between impedance and admittance controllers. In the admittance controller the increment of the index (I) is a function of the external torque disturbance T_{dp}. The ankle kinematics data started and finished in the middle of the swing phase (vector indices I_0 and I_f , respectively). The index for the data at the beginning of the foot-flat phase (index I_{ff}), and the expected heel-strike (index I_{hs}) was known. These points were important as the finite-state machine was programmed to switch from impedance to admittance controller at heel-strike and from admittance to impedance controller at the beginning of the foot-flat phase.

B. Impedance Controller

The robot has two DC motors working together to generate torques in 2 DOFs. When the motors rotate in opposite directions DP motion is generated. When they move in the same direction IE motion is generated. This requires two controllers, one for each motor, to be used. The impedance controllers (Fig. 3) use position encoders mounted in their respective gear boxes to determine the position of the foot. The controllers use the desired and feedback positions to derive the actuators desired torques using a PD controller. The torque feedback is obtained from the WB. The torque feedback gain K adjusts the quasi-static stiffness of the ankle, which will be discussed later. The desired torque and torque feedback are then used to derive the appropriate control input to the motors using a PD controller.

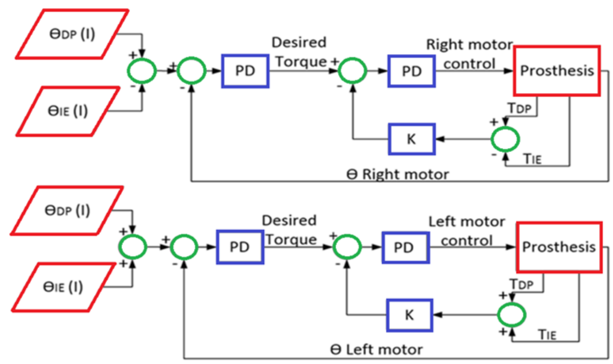

Figure 3. Impedance controllers for the left and right motors. The reference angle for the left motor controller is the sum of the DP and IE angles, while for the right motor controller is the difference between DP and IE angles. The torque feedback in the left motor controller is the sum of the DP and IE torques (T_{DP} and T_{IE} respectivelly), while for the right motor controller is the difference between DP and IE torques. This is necessary since the outputs (both angle and torque) of the prosthesis in DP are proportional to the output of both motors, and the outputs in IE are the difference between the output of the motors.

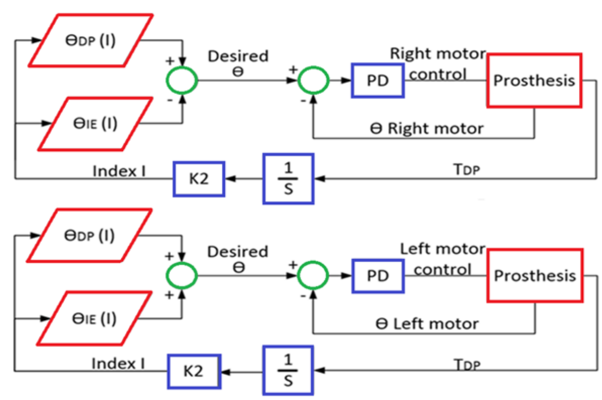

Figure 4. Admittance controllers for the left and right motors. The reference angle for the left motor controller is the sum of the DP and IE angles, while for the right motor controller it is the difference between DP and IE angles. The admittance comptroller uses the torque feedback in DP (T_{DP}) to update the lookup table index I.

C. Admittance Controller

An admittance controller requires torque feedback to update an inner position controller. The proposed admittance controller was designed to use the look-up table of ankle angles to update the inner position control (Fig. 4). The controller integrates the ground reaction torque feedback (in DP) and uses the result as the index of the look-up table of the ankle angles, which are then used as the reference input to an internal position controller. This way, an external torque input will make the prosthesis to advance through the gait cycle and the recorded data with speed proportional to the external torque, while the absence of an external torque will keep the foot stationary. This allows the foot to follow the prerecorded angular trajectories while admitting external torque to produce motion. The foot will start and stop moving automatically based on the detection of heel-strike.

V. Preliminary Evaluation Experiments

A. Impedance Controller in Quasi-static Condition

An experiment was conducted to study the ability of the impedance controller to change the quasi-static stiffness of the prosthesis. The prosthesis was attached to an Anklebot, a lower extremity therapeutic robot (Interactive Motion Technologies), which is capable of applying torques to the ankle in both frontal and sagittal planes, and to record the resultant angles. The attachment of the Anklebot to the prosthesis was identical to the procedure for attaching the Anklebot to the human foot, as described in [9]. Six tests were performed in each plane, where the prosthesis impedance controller was set at a reference angle of zero degrees in both DP and IE. In each test the torque feedback gain K was changed from values ranging from -0.5 to 1.5. In the DP tests, the Anklebot moved the foot from the equilibrium point to 4° dorsiflexion and followed by moving the foot to 4° plantarflexion. In the IE test, the Anklebot moved the foot to 4° eversion from the equilibrium point and in a continuous motion returned the foot to 4° inversion. The movement speed was set to 5°/second and the data at the encoders were recorded at a sampling rate of 200 samples per second. The results were filtered with 0.5 Hz cutoff frequency to remove sensor noise.

The results of the tests in DP with different gains are shown in Fig. 5, depicting the unloading, transition, and loading phases of the ankle. It can be seen that the change in the feedback gain effectively changed the stiffness of the ankle (or the slope of torque-angle curve in Fig. 5). Zero gain caused the prosthesis to behave as a passive prosthesis, since the prosthesis is not a backdrivable mechanism. Negative gains caused the prosthesis stiffness to increase compared to the zero gain test. Positive gains resulted in a decrease in the prosthesis stiffness compared to the zero gain case. All the gains produced near linear changes in DP torque with respect to the change in angle with average Pearson Product-Moment Correlation Coefficients (PPMCC), which is a measure of the linear correlation, of 0.99±0.01. Some deviation was observed near the origin and was caused by the transition in the ankle from loading to unloading and its effects on the bending of the composite plate. Straight lines of best fit using the least square method were fitted to each of the tests in Fig 5, and the slopes (stiffness of the ankle) of these lines were plotted against their respective gains in Fig. 6. The PPMC between the torque feedback gain and the quasi-static stiffness of the prosthesis was -0.99, showing that there is a near negative linear correlation between the change in torque feedback gain and the quasi-static stiffness of the prosthesis. The stiffness of the prosthesis in DP was found to be 2.09 Nm/degree with a -0.5 gain that decreased to 0.92 Nm/degree at gain 1.5.

The results of the test in IE can be seen in Fig. 7, where the change in the feedback gain effectively changed the stiffness of the ankle in IE, similarly to the test in DP, with average PPMCC of 0.99+/-0.002. The quasi-static stiffness of the ankle was plotted against their respective gains, as shown in Fig. 8 and had a PPMCC of -0.96. The prosthesis stiffness in IE was found to be 0.53 Nm/degree at a -0.5 gain that decreased to 0.17 Nm/degree at gain 1.5.

B. Preliminary Controller Evaluation Using a Circular Treadmill Evaluation Platform

During the development of assistive robots, it is necessary to evaluate their performance. It is a challenging task, especially considering that in a development stage, the devices are not in a safe or suitable condition to perform experiments with amputees. To overcome this issue, a circular treadmill was developed (Fig. 9), allowing the ankle-foot prosthesis to be evaluated during walk in a turning pattern without the need of human subjects. Additionally, the platform provides a consistent environment with repeatable experiments.

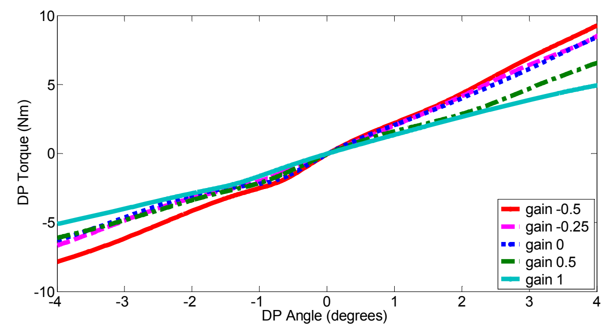

Figure 5. DP torque-angle relationship in the prosthesis with impedance control and different torque feedback gains. Negative angles are plantarflexion, positive angles are dorsiflexion.

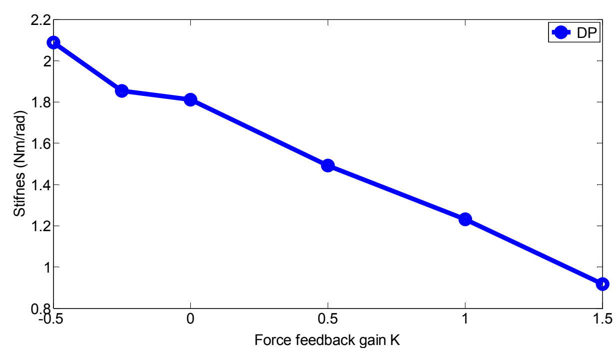

Figure 6. Ankle stiffness in DP at different DP torque feedback gains.

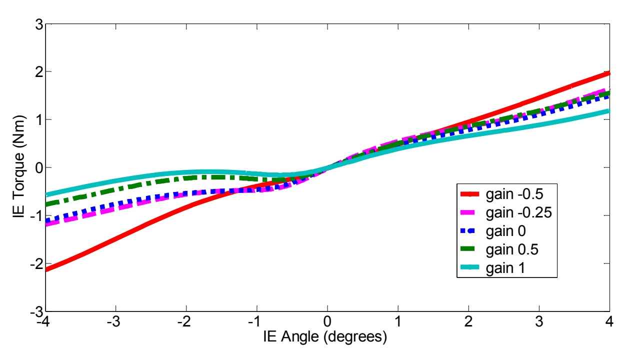

Figure 7. IE torque-angle relationship in the prosthesis with impedance control and different torque feedback gains. Negative angles are plantarflexion, positive angles are dorsiflexion.

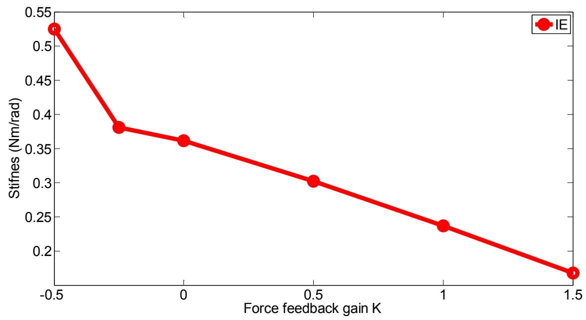

Figure 8. Ankle stiffness in IE at different IE torque feedback gains.

The platform can lift and lower the foot and apply weight to emulate a human walk. The radius of the turn of each step can be increased or decreased by sliding the frame (I) so the foot is closer to or farther away from the center of the treadmill. Also, the weight supported by the prosthetic leg can be controlled by adding or removing weights or by sliding the joint (E) closer to or farther away from the weight (H). The speed of the treadmill disk is controlled using an open loop controller that can provide a maximum walking speed of 1.63 m/sec. The lifting mechanism uses a PD position controller were the input is a sine wave with the same frequency as the gait (it is synchronized with the finite-state machine). The amplitude and time shift are dependent on the ankle-foot prosthetic tuning, amount of weight being used, and the position of the prosthesis with respect to the frame and treadmill. The lifting mechanism is capable of lifting 118 kg at 10.6 m/s. At this point the prosthesis shank angle is not controlled, and the “knee” is a passive 2-DOF joint resulting in a free swing in the forward phase. Future designs will incorporate an active knee joint to allow the control of the swing phase.

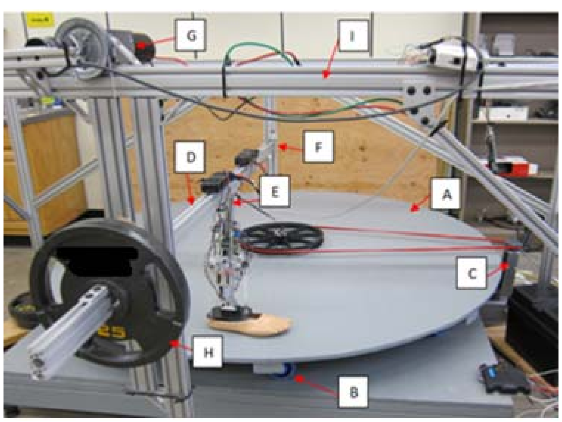

Figure 9. Circular treadmill and its main components. The circular treadmill is composed of a wooden disk with a 1m radius (A). 8 coaster wheels (B) are connected to the the disk for weight bearing, and a heavy-duty turn table (not visible) is attached in the center and under the disk for both weight-bearing and constraining the disk from sliding on the horizontal plane. A motor and planetary gear box (C) powers the rotation of the disk. The prosthetic robot is connected to a horizontal bar (D) by a universal joint (E) which acts as a passive knee. The bar has one end coupled to a pivot (F) and the other end attached to a cable connecting it to a motor and gear box (G) which can raise and lower the bar and the robot. The same end of the bar is coupled to a weight (H) which is supported by the prosthetic leg when the weight is lowered or by the motor and gear box (G) when the bar is raised. The prosthetic leg, bar, motor and gear box, and weight are attached to an aluminum frame (I) which is not coupled to the treadmill except when the foot contacts the wooden disk.

C. Prosthesis Test on the Treadmill

The circular treadmill was used to test the prosthesis performance with the finite state machine and compare the results with the performance of the device using a position control and no control (passive prosthesis). The impedance/admittance controllers were set with a torque feedback gain of 0.5 for both DP and IE. The horizontal bar (D) was subjected to 11.4 kgf static load (H). Due to the bar mechanism, the load on the foot was equivalent to 22.8 kgf. The radius of turn during the walk was set to 0.85 m.

During the tests the ground reaction forces were obtained from the WBs and used to estimate the resultant torque applied to the ankle (Figs. 10 and 11). It was seen that during the swing phase there were zero torque feedbacks since the foot was not contacting the ground. When contact happened, the passive prosthesis showed the largest reaction torques which saturated the data acquisition system at 15 Nm torque. The position controller decreased the DP torque at heel-strike, but showed similar torque at push-off when compared to the passive prosthesis. The impedance/admittance control showed the least amount of DP torques at both heel-strike and push-off. IE torques were the largest in the passive prosthesis and the impedance/admittance controller showed the least amount of torque. Inversion torques were larger than eversion torques for all experiments, which is expected since the foot is turning left as it walks on the treadmill, putting pressure on the inside edge of the foot.

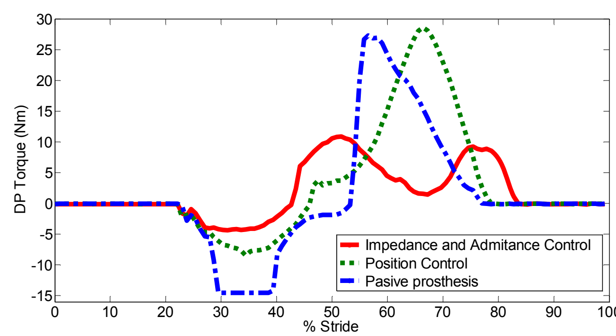

Figure 10. Ankle external torque in DP during a representative gait cycle with different control strategies. Negative torques induce plantarflexion (heel-strike) and positive torques induce dorsiflexion (push-off). With the passive prosthesis, the data acquisition system saturates during heel-strike at -15 Nm torque.

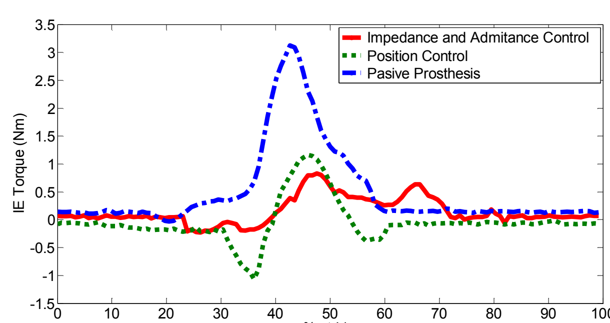

Figure 11. Ankle external torque in IE during a representative gait cycle with different control strategies. Negative torques induce eversion and positive torques induce inversion.

It is seen in Figs. 10 and 11 that the impedance/admittance controller was capable of reducing the amount of external torque in the foot in both frontal and sagittal planes; however, it increased the amount of time the foot was in contact with the ground. This is expected since the impedance controller is effectively changing the stiffness of the ankle by applying a torque in the same direction as the disturbance torque. This causes the foot to be at a larger dorsiflexion angle compared to the reference input, resulting in an extended time for push-off. Smaller reaction forces at heel-strike are desirable, as these forces are directly transferred to the user. At push-off, smaller forces may reduce the energy consumption of the prosthesis. However, the prosthesis needs to have enough stiffness at heel-strike to control the impact and generate enough torque for forward propulsion during push-off. The required stiffness of the ankle at different phases of the gait will be addressed in our future work. The impedance/admittance control was capable of reducing the external reaction torques, but this is only desirable if the torques are large enough to follow the desired trajectory.

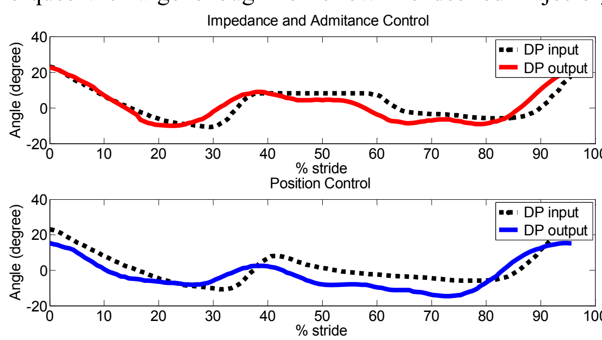

Figure 12. Input and output (compensated for 75 milliseconds time delay) of the ankle trajectory in DP during a representative gait cycle with admittance and impedance control.

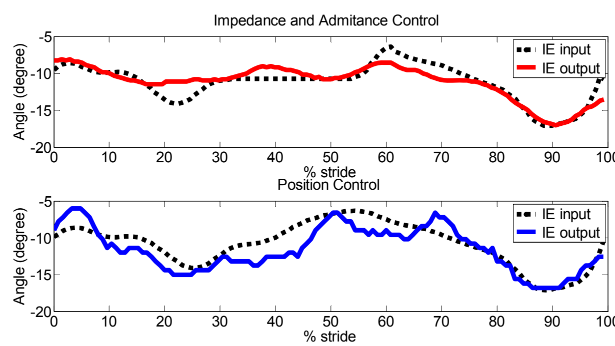

Figure 13. Input and output (compensated for 75 milliseconds time delay) of the ankle trajectory in IE during a representative gait cycle with admittance and impedance control.

The input and output trajectories of the foot in both DP and IE during the tests can be seen in Figs. 12 and 13, respectively. The input data is the time history of rotations of a human ankle during gait, and the output is the trajectory of the ankle obtained from the quadrature encoders in the prosthesis. For ease of comparison, the output plots have a time shift to remove the 75 milliseconds delay of the output. This delay is resultant from the motors and gear boxes and will be improved in the next generation of the prosthesis. From Fig. 12, it can be seen that the impedance/admittance controller input held the ankle constant from near 35% up to 60% of the stride, due to the state machine reaching the index of the expected heel-strike, but heel-strike has not happened yet. The tracking performance was better in DP when compared to the tracking in IE (Fig. 13). Due to the physical characteristics of the prosthesis, small angular differences between the left and right motors caused larger changes in the foot rotations in IE, making the system more sensitive to disturbances and noise when compared to DP.

VI. Conclusion

In this paper, we presented the preliminary steps towards the development of a finite state machine for control of a two degree of freedom ankle-foot prosthesis. Strain gauges were installed on the prosthesis' foot and were used to estimate the external torques in the frontal and sagittal planes. The estimated torques were used for torque feedback in impedance and admittance controllers. The finite state machine was capable of switching between impedance and admittance controllers based on the state of the gait. The quasi-static stiffness of the prosthesis in impedance controller was evaluated, showing a near linear relationship between the torque feedback gain and the quasi-static stiffness of the ankle. The finite state machine and controllers were also evaluated with a circular treadmill evaluation platform, and the results were compared to the performance of the prosthesis with a position controller and with no control at all. These experiments showed that the impedance/admittance controller was capable of tracking the desired reference trajectory while decreasing the reaction torques at the foot.

Acknowledgment

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