Proceedings of the ASME 2014 Dynamic Systems and Control Conference
DSCC2014
October 22-24, 2014, San Antonio, TX, USA

DSCC2014-6032

IMPEDANCE AND ADMITTANCE CONTROLLER FOR A MULTI-AXIS POWERED ANKLE-FOOT PROSTHESIS

Evandro M. Ficanha
emficanh@mtu.edu

Mohammad Rastgaar
rastgaar@mtu.edu

Department of Mechanical Engineering-Engineering Mechanics
Michigan Technological University, Houghton, Michigan 49931

ABSTRACT

This paper introduces a finite state machine to select between impedance and admittance control for a powered ankle-foot prosthesis controllable in both Dorsiflexion-Plantarflexion (DP) and Inversion-Eversion (IE). Strain gauges are installed on the prosthesis' foot to measure the strain caused by ground reaction forces, which are correlated to the external torques in DP and IE. The external torques are used for the admittance and impedance controllers. Additionally, the finite state machine uses the strain gauges feedback to detect the heel-strike and switch to admittance control. The admittance control accepts torque feedback to generate motion, this way larger feedback torques effectively reduces the stiffness of the ankle. During push off, the finite state machine switches to impedance control, accepting motion feedback to generate the appropriated torques. The quasi-static stiffness of the prosthesis with impedance control was tested, showing a near linear relationship between the torque feedback gain and the stiffness of the ankle. The finite state machine and controllers were also evaluated using a custom-made circular treadmill and the results were compared to the results of position and passive controllers; showing that the impedance/admittance controller was capable of tracking the desired input trajectory while decreasing the required torque at the ankle joint.

INTRODUCTION

Walking in a straight line requires a complex modulation of muscle contractions to control the ankle's stiffness and generate forward propulsion. Similar muscle contractions are required to generate the appropriate ground reaction forces to steer the body while turning [1]. Below knee amputees with passive prosthesis expend 20-30% more energy than non-amputees to walk at the same speed, resulting in a preferred walking speed which is 30-40% slower than non amputees [2, 3]. As a possible solution, powered prostheses have been developed and it is shown that they reduce the metabolic cost during straight walking by providing energy to the gait at push-off [4, 5]. While the focus on developing powered prostheses has been on increased mobility in forward locomotion; it has been shown that daily activities contain an average of 25% turning steps [6]. Turning requires modulation of the ankles impedance in both DP and IE directions to control the lateral and forward reaction forces to maintain the body's center of mass along the desired trajectory; resulting in increased lateral and forward forces when compared to the straight walking [7]. Due to the lack of appropriate propulsion from their passive prostheses, amputees rely on different gait strategies than non amputees [1]; suggesting that amputees can benefit from powered prostheses capable of providing power in both DP and IE with impedance modulation similar to the human ankle.

While physical systems interact with each other, they are behaving 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) [8]. The coupled mechanical systems must physically complement each other, meaning that in any degree of freedom, if one system is an impedance, the other system has to be an admittance [8]. During gait, at the moment the heel interacts with the ground (heel-strike) the ankle is being manipulated by the environment since the ankle accepts the external force and generates the appropriate motion, so it may be considered as a system in admittance. At push off, the ankle manipulates the environment, generating the necessary torques to produce the required motion, and so it may be considered as impedance. This suggests that ankle-foot prostheses should use an admittance controller at heel -strike and an impedance controller at push off. In general, an impedance controller uses position encoders mounted in the actuators to determine the position of the robot end effector. The controller uses the desired and feedback positions to generate the actuators' desired torques. Torque sensors are used to provide the means for estimation of the external torque feedback that, along

 Copyright © 2014 by ASME

with the desired torque, are used to define the appropriated input to the motors [9]. An admittance controller or 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 [9]. In other words, the impedance controller accepts external motion inputs and generates output torques, while the admittance controller accepts external torque inputs and generates output motions.

The mechanical impedance of the human ankle has been studied by many researchers [10-15]; also the quasi-static stiffness of the ankle has been measured [16]. Quasi-static stiffness of the ankle in the sagittal plane has been used in the design of ankle-foot prostheses capable of producing a positive work during the gait. Sup et al. developed a knee and ankle prosthesis capable of controlling the impedance of both the knee and ankle joints in the sagittal plane by controlling the neutral position of the foot during the gait [17-20]. BiOM is a commercially available ankle prosthesis that is capable of providing the necessary energy during push-off (plantarflexion); therefore, actively contributes in gait and lowers the gait metabolic cost by 8.9% to 12.1% at different gait speeds compared to a passive prosthesis [21]. The controller in both of the aforementioned prostheses use a finite state machine to identify the gait phase and estimate the appropriate ankle torques. This allows the prostheses to adapt to different gaits scenarios such as different cadence or walking uphill and downhill [22]. Although the aforementioned prostheses improve the gait of amputees, they are designed to modulate the ankle torques in the sagittal plane only.

In this paper the concept and prototype of a multi-axis powered ankle-foot prosthesis capable of controlling two degrees of freedom is presented. Additionally, the paper describes the method used to measure the torque feedback in both DP and IE using strain gauges; the preliminary steps toward development of impedance and admittance controllers based on the torques and positions feedback; the development of a finite state machine to identify the state of the gait and switch between admittance and impedance controllers, and preliminary tests to evaluate the performance of the prosthesis while walking on a circular treadmill as an evaluation platform.

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

A multi-axis ankle prosthesis may enhance gait efficiency by extending the control of IE during walking in a straight line and turning. This novel design is aimed to enable the device to adapt to uneven and inclined ground surfaces and allow the amputees to benefit more from their prostheses rather than using their hip joint as a compensatory gait mechanism; enabling a more agile and natural gait with less stress on other joints.

A prototype design of a cable-driven ankle-foot prosthesis controllable in both DP and IE was designed and fabricated in an effort to study the feasibility of the steering and maneuverability

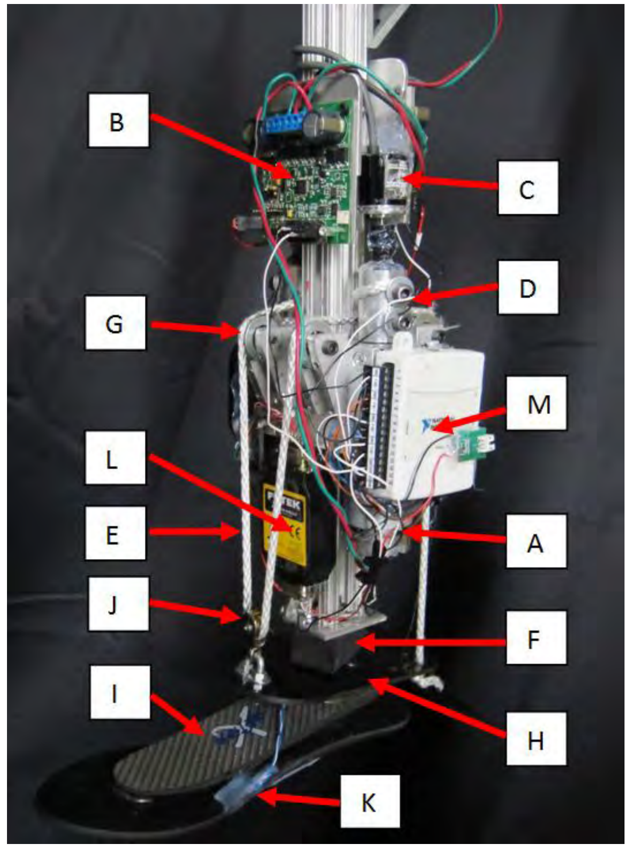


FIGURE 1: Two-DOF ankle-foot prosthesis prototype.

with a 2 degrees of freedom (DOF) ankle joint (Fig. 1). The design allowed for the range of motion (ROM) and angular velocity similar to the human ankle during straight walking and turning while producing enough torque for propulsion.

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 estimated from ground reaction forces provided by six strain gauges in the foot (K and Fig. 2) 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. The mechanism is capable of moving in DP when the motors rotate in opposite directions and in IE when the motors rotate in the same direction. Also, any combination of DP and IE can be

Copyright © 2014 by ASME

obtained by combining different amounts of rotations in each motor.

ANKLE TORQUE AND ANGLE FEEDBACK

To develop impedance and admittance controllers, 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.

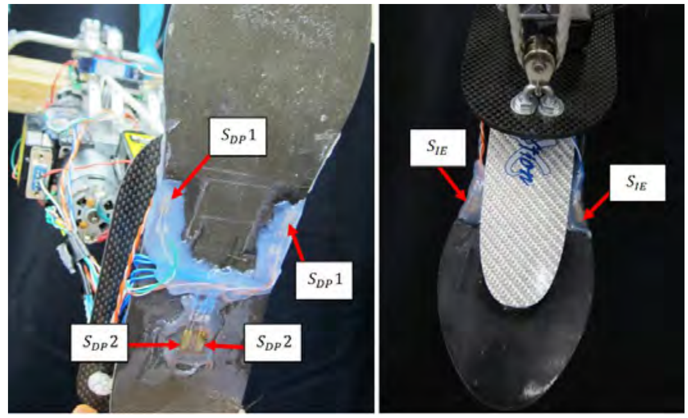
$$\theta_{DP} = K_{dp}\left(\frac{\theta_{Left} + \theta_{Right}}{2}\right) \tag{1}$$

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

The ground reaction torques were estimated using strain gauges, as they are commonly used on load cells to measure the strains of structures due to external forces. The strain gauges change the resistance as they stretch or contract and their change in resistance can be correlated to the strain of the object they are attached to and consequentially the force or torque applied to the object. Strain gauges are typically wired in a Wheatstone Bridge configuration using four strain gauges (some of the strain gauges can be replaced by resistors). The increase or decrease in resistance of two of the strain gauges (placed in opposite sides of the bridge) causes the output of the bridge to increase or decrease, respectively. The opposite is also true for the other two strain gauges where the increase or decrease in resistance decreases or increases the output voltage of the bridge, respectively.

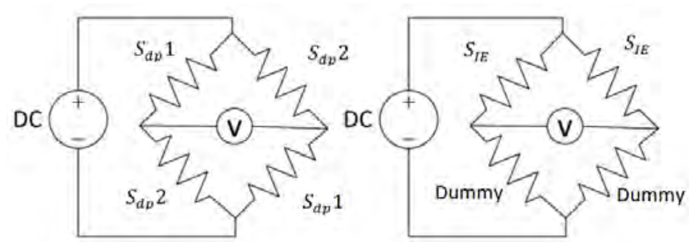
For estimating the torque in DP, four strain gauges were attached to the sole of the foot (Fig. 2-A). The strain gauges $S_{dp}2$ were located behind the center of rotation of the ankle in DP and were wired into opposite sides of the Wheatstone Bridge (Fig. 3-A). Any ground reaction force at the hill caused a decrease in voltage of the Wheatstone Bridge, which could be correlated to the torque in DP of the foot when the heel was interacting with the ground (e.g. heel-strike). The strain gauges $S_{dp}1$ were located in front of the center of rotation of the ankle in DP and wired in opposite sides of the Wheatstone Bridge (Fig. 3-A). Any ground reaction force from the ground at the front of the foot caused an increase in the output voltage of the Wheatstone Bridge, which could be correlated to the torque in DP of the foot when it was contacting the ground (e.g. push off). Note that when the foot was flat on the ground, the output from the strain gauges $S_{dp}1$ cancel the output from the strain gauges $S_{dp}2$; therefore, the resultant voltage could always be correlated to the net DP torque in the ankle.

For estimating the torque in IE, two strain gauges were attached to the top of the foot as seen in Fig. 2-B. The other two strain gauges were attached to an inert piece of carbon fiber, but



(A) **(B)**

FIGURE 2: Strain gauge placement in the prosthesis. A: For DP torque two strain gauges were used for push-off torque estimation ($S_{DP}1$) and two strain gauges were used for heel-strike torque estimation ($S_{DP}2$). B: For IE torque two strain gauges are used for push-off torque estimation ($S_{IE}1$).



(A) (B)

FIGURE 3: A: Strain gauge placement in the Wheatstone Bridge for DP torque estimation. Two strain gauges were used for push-off DP torque estimation ($S_{DP}1$) and two strain gauges were used for DP heel-strike torque estimation ($S_{DP}2$). B: Strain gauge placement in the Wheatstone Bridge for IE torque estimation. Two strain gauges are used for push-off IE torque estimation (S_{IE}).

could also be replaced by two resistors with resistance identical to the strain gauges. The strain gauges were placed on the outside edges of the foot and were on the same side of the Wheatstone Bridge (Fig. 3); hence, the difference in strains caused an increase or decrease in voltage at the bridge. 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 Bridge insensitive to the torque in DP, since the strain gauges were in the opposite ends of the bridge. Therefore, if they both contract or stretch by the same amount as it would happen in the presence of a DP torque, the output would not be affected. This feature is important since it is necessary to have the IE torque estimation to be decoupled from the DP torque.

To correlate the strain gauge readings to the actual disturbance torques, a Kistler® Type 5233A force plate was used to measure the external force applied during static loading tests. The tests consisted of loading the foot in different configurations and recording the applied force and the corresponding strain

Copyright © 2014 by ASME

measurement. The tests were: plantarflexion by applying a load when the heel was in contact with the ground, dorsiflexion by applying a load when the forefoot was in contact with the ground, eversion by applying a load when the right edge of the forefoot was in contact with the ground, and inversion by applying a load when the left edge of the forefoot was in contact with the ground. From the external forces, the geometry of the foot, and the strain measurements, the applied torques were calculated. It is important to note that for DP, the proportional factor between the external force and the strains measured at heel loading and forefoot loading were not the same, since the strain gauges were attached to two different areas of the prosthetic foot. The proportional factors for the strain gauges at heel loading and forefoot loading were estimated as 1.41 Nm/volt and 19.52 Nm/volt, respectively. As a result, different proportional factors needed to be used depending if the strain measurement was positive or negative. For IE, the proportional factors for inversion and eversion torques were closer (4.43 Nm/volt and 3.55 Nm/volt, respectively), which was expected since the foot is near symmetrical about its sagittal plane.

CONTROLLER
Finite State Machine

A finite-state machine was designed to switch between impedance and admittance controller (Fig. 4). The finite-state machine uses pre-recorded time-history of the ankle angles in DP and IE during normal walk and real time torque feedback estimated from the strain gauges to switch between the states. The recorded ankle angles of an unimpaired human subject were measured using a motion capture camera system. It is aimed to use the trajectories of the ankle of unimpaired human beings as a start point for a powered prosthesis, which can be tuned to the specific needs of amputees. The ankle angles were accessible as look-up data table to the estate machine and controllers. The vectors with the ankle data started and finished in the middle of the swing phase (vector indices I_0 and I_f , respectively). Also the index for the data at the beginning of the flat foot (index I_{ff}), and expected heel-strike (index I_{hs}) were known. These points were important as the finite-state machine needed to switch from impedance controller to admittance controller at heel-strike and from admittance controller to impedance controller at the initiation of the flat foot phase.

From Fig. 4 it can be seen that the foot starts at the middle of the swing phase and moves with the active 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 angle at the beginning of the heel-strike phase and will hold that position until heel strike is detected. This is important as the robot can adjust to small variations in walking speed, and will start and stop automatically when the gait starts or stops. Once the foot reaches flat foot, the control switches back to an impedance

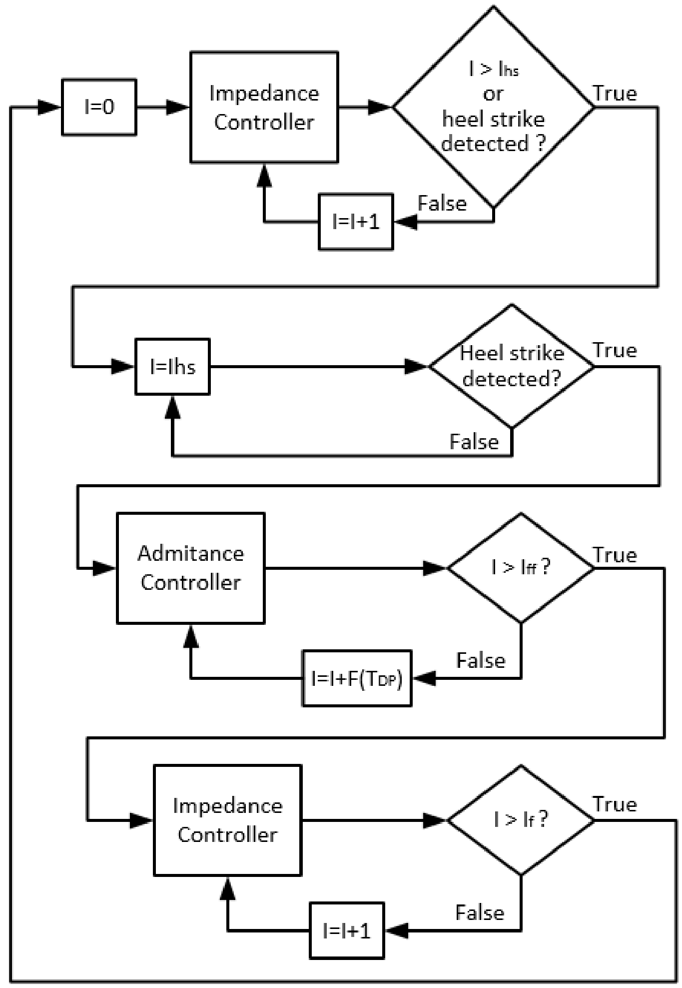


FIGURE 4: Finite-state machine to switch between impedance and admittance controllers. In the admittance controller the increment of the index (I) is a function of the external torque disturbance T_{dp}.

controller until the foot reaches the middle of the swing phase where the index I is reset to zero and the cycle 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.

Impedance Controller

The robot has two DC motors working together to produce torques that move the foot with respect to the pylon. When the motors rotate in opposite directions DP motion is produced. When they move in the same direction IE motion is produced. This implies that two controllers are needed, one for each motor.

The impedance controller (Fig. 5) for both the left and right motors uses position encoders mounted in their respective gear boxes to determine the position of the foot. The controller uses the desired and feedback positions to derive the actuators

Copyright © 2014 by ASME

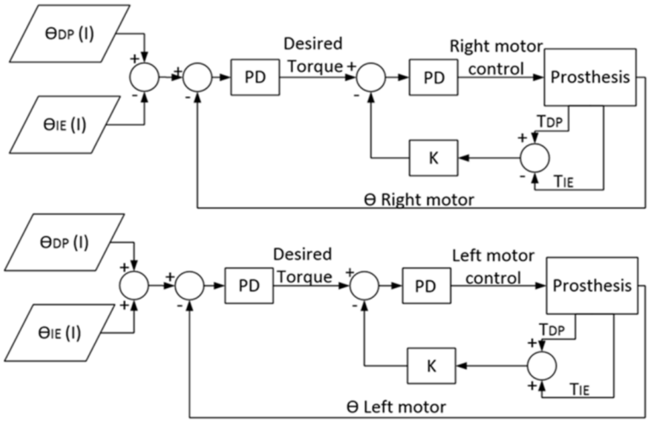


FIGURE 5: Impedance controllers for the left and right motors. The reference angle for the left motor controller is the sum of the DP and IE angles, while for the right motor controller is the difference between DP and IE angles. The torque feedback in the left motor controller is the sum of the DP and IE torques, while for the right motor controller is the difference between DP and IE torques.

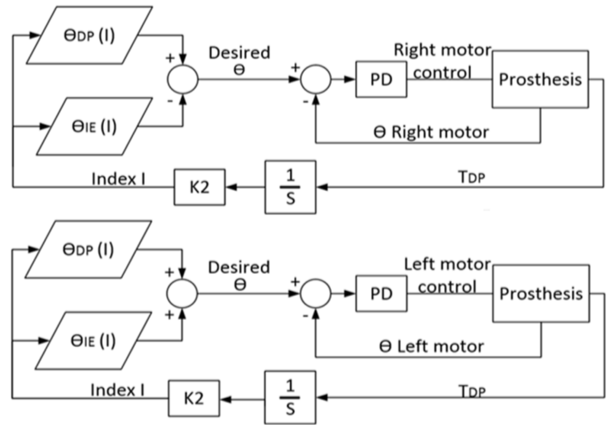


FIGURE 6: Admittance controllers for the left and right motors. The reference angle for the left motor controller is the sum of the DP and IE angles, while for the right motor controller is the difference between DP and IE angles. The admittance comptroller uses the torque feedback in DP to update the lookup table index I.

desired torques using a PD controller. The torque feedback calculated from the strain gauges are used to estimate the ground reaction torques to be used as the feedback. The torque feedback gain K adjusts the quasi-static stiffness of the ankle, which will be explored later. The desired torque and torque feedback are then used to derive the appropriate control input to the motors using a PD controller. Note that the reference angle for the left motor controller (looking at the prosthesis from the front) is the sum of the DP and IE angles, while the reference angle for the right motor controller is the difference between DP and IE angles. Similarly, the torque feedback for the left motor controller is the sum of the DP and IE ground reaction torques, while the feedback torque for the right motor controller is the difference between DP and IE environment torques. This is necessary since the outputs (both angle in 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.

Admittance Controller

An admittance controller requires torque feedback to update an inner position controller. The proposed admittance controller was designed to use a look-up data table for updating the inner position control (Fig. 6). The control integrates the ground reaction torque feedback (in DP) to increase the index of the look-up table of the ankle angles proportionally to the external torque. This way, an external torque input will make the prosthesis able to advance through the data vector, 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. Also, at heel-strike the foot will not move unless it contacts the ground; therefore, the foot will start and stop moving automatically based on the external torque feedback. It is important to note that this controller will only engage when the motion of the device is known, and the external torque is only used to control how fast the robot will follow the predetermined trajectory. At this point the prosthesis does not have IE torque feedback at the heel, so only the DP torque was used to update the controller at heel strike.

PRELIMINARY EVALUATION EXPERIMENTS

Impedance Controller in Quasi-static Condition

To evaluate the impedance controller and its ability to change the quasi-static stiffness of the ankle, an experiment was designed to record the quasi-static torque-angle relationship of the prosthesis. The prosthesis was attached to an Anklebot, a lower extremity therapeutic robot (Interactive Motion Technologies, inv.) as seen in Fig. 7. The Anklebot is capable of applying torques and record angular motion of the ankle in both DP and IE, this makes it suitable for the experiment. To test the DP stiffness, the prosthesis impedance controller was set at a reference angle of zero degrees and a constant torque feedback gain K for each test. Six tests were performed setting the gain K at different values ranging from -0.5 to 1.5. In each test the Anklebot moved the foot from the equilibrium point to 6° dorsiflexion and followed by moving the foot to 6° plantarflexion. 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 the sensor noise.

The results of the tests with different gains are shown in Fig. 8, 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 in DP (or the slope of torque-angle curve in Fig. 8). Zero gain caused the prosthesis to behave as a passive prosthesis, since it is not a backdrivable mechanism. Negative gains caused the prosthesis stiffness to

Copyright © 2014 by ASME

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 some deviation near the origin caused by the transition in the ankle from loading to unloading and its effects on the bending of the composite plate. Best fit lines were fit to each of the tests in Fig 8, and the slopes (stiffness of the ankle) of these lines were plotted against their respective gains in Fig. 9. It can be seen that there is a near linear relation between the change in torque feedback gain and the quasi-static stiffness of the prosthesis with positive gains. 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.
To test the IE stiffness a similar experiment was conducted. Six tests were performed with the same torque feedback gains as the DP test. In each test the Anklebot moved the foot to 12° eversion from the equilibrium point and in a continuous motion returned the foot to 12° inversion. Large angular displacements were needed in the IE test since, by design, the ankle-foot prosthesis shows a smaller passive stiffness in IE than in DP. It can be seen in Fig. 10 that the change in the feedback gain effectively

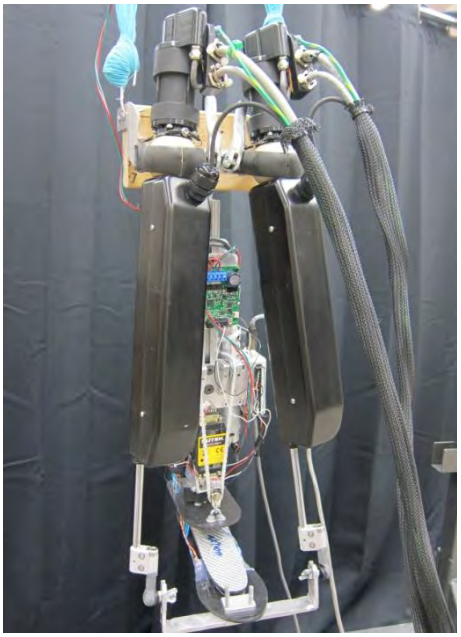


FIGURE 7: Ankle-foot prosthesis attached to the Anklebot.

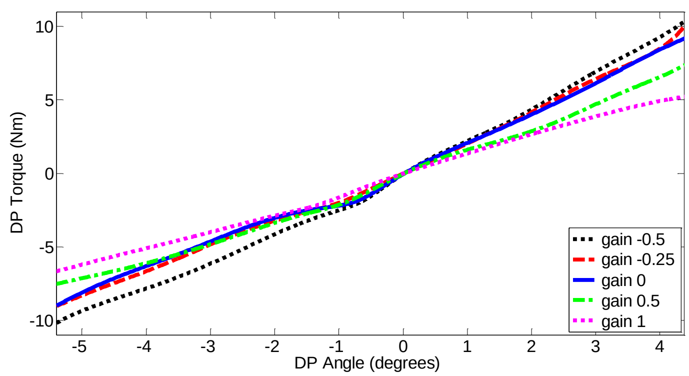


FIGURE 8: DP torque-angle relationship in the prosthesis with impedance control and different torque feedback gains. Negative angles are plantarflexion, positive angles are dorsiflexion.

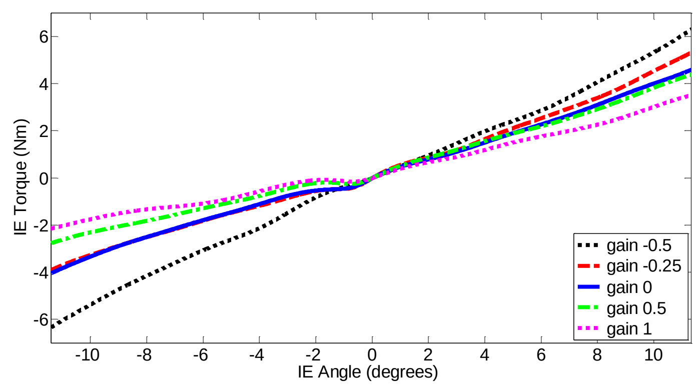


FIGURE 10: IE torque-angle relationship in the prosthesis with impedance control and different torque feedback gains. Negative angles are plantarflexion, positive angles are dorsiflexion.

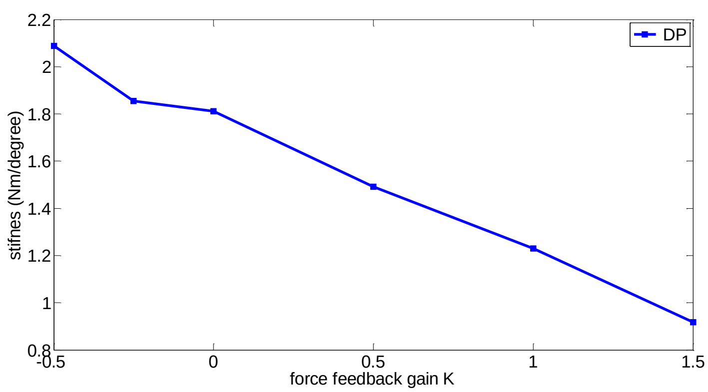


FIGURE 9: Ankle stiffness in DP at different DP torque feedback gains.

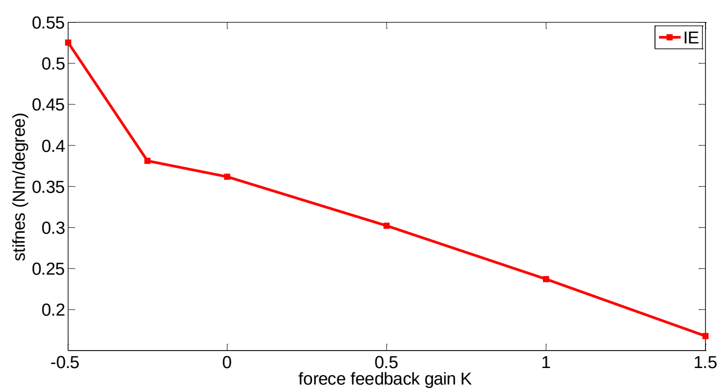


FIGURE 11: Ankle stiffness in DP at different DP torque feedback gains.

Copyright © 2014 by ASME

changed the stiffness of the ankle in IE. Similar to the DP test, negative gains caused the prosthesis stiffness to increase compared to the zero gain test. With positive gains, the prosthesis stiffness decreased compared to the zero gain test. All the gains produced near linear changes in IE torque with respect to the change in angle, with some deviation near the origin caused by the transition in the ankle from loading to unloading. Similar to DP tests, the quasi-static stiffness of the ankle were plotted against their respective gains, as shown in Fig. 11, indicating a near linear relation between the change in torque feedback gain and the quasi-static stiffness of the prosthesis for the positive gain. 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.

Preliminary Controller Evaluation in Dynamic Condition Using a Circular Treadmill Evaluation Platform

It is of interest to evaluate the performance of the prosthesis and controller during gait. One of the challenges of testing prosthetic devices is that usually a human subject needs to be using them, making the tests non-repeatable and inconsistent. To evaluate the ankle-foot robot, a circular treadmill was developed (Fig. 12), allowing the ankle-foot robot to be examined during walk in a turning pattern without the need of human interaction. The circular treadmill is composed of a wooden disk with a 1m radius (A). 8 coaster wheels (B) are connected to the outside lower edge of the disk for weight bearing, and a heavy-duty turn table (not visible) is connected in the center 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 connected to a pivot (F) and the other end connected to a cable connected to a motor and gear box (G) which can raise and lower the bar and the robot. The same end of the bar is connected 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 at the time of the stance when the foot contacts the wooden disk.

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 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. The final gear ratio is 341/1 resulting in a maximum walking speed of 1.63 m/sec which is sufficient considering the average preferred human walking speeds for young adults is 1.30 m/s [23]. The lifting mechanism uses a PD

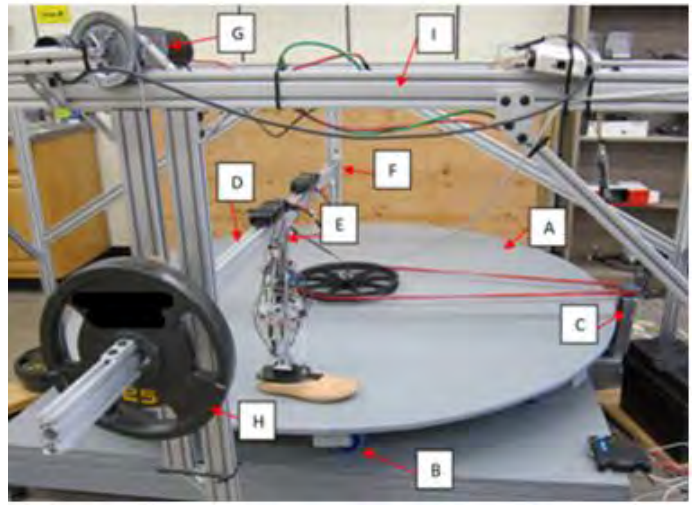


FIGURE 12: Circular treadmill and its main components.

(A) (B) (C)

FIGURE 13: Circular treadmill and prosthetic ankle foot robot. A- Heel-strike. B- Flat foot. C- Push off.

controller with feedback from a quadrature encoder and 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 prosthetic ankle-foot 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, although the weight supported by the prosthetic leg is higher and depends on the position of the shank of the robot with respect to the beam (D). At this point the prosthesis shank angle is not controlled, and the “knee” joint is a passive one DOF joint resulting in a free swing forward phase. Future designs will incorporate an active knee joint to allow a controlled swing phase speed. Fig. 13 shows the foot at different states of the gait while walking on the treadmill.

Prosthesis Test on the Treadmill

The circular treadmill was used to test the prosthesis performance with the impedance/admittance control and compare the results with the performance of the device using a position control and no control (passive prosthesis). The impedance/admittance controller was set with a torque feedback gain 0.5 for both DP and IE. As shown in Fig. 12, 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.

Copyright © 2014 by ASME

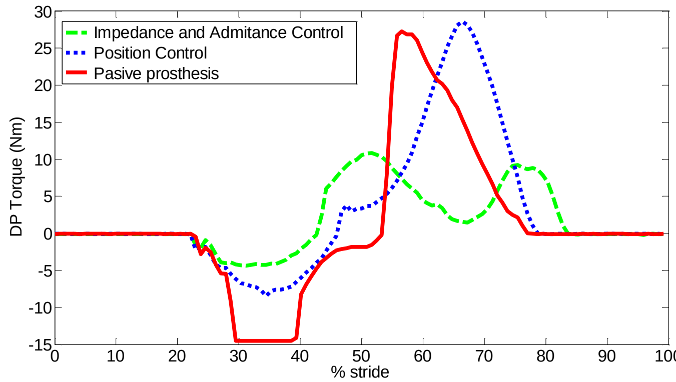


FIGURE 14: Ankle external torque in DP during a representative gait cycle with different control strategies. Negative torques induces plantarflexion (heel-strike) and positive torques induces dorsiflexion (push off). With the passive prosthesis, the data acquisition system saturates during heel-strike at -15 Nm torque.

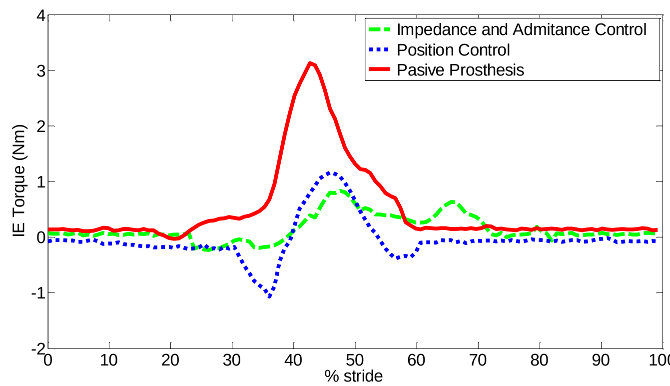


FIGURE 15: Ankle external torque in IE during a representative gait cycle with different control strategies. Negative torques induces eversion and positive torques induces inversion.

The radius of turn during the walk was set to 0.85 m. The position controller used a PD controller to follow the trajectory of the previously recorded data of a human subject ankle in both DP and IE. With all the controls off, the device behaved as a passive prosthesis with the stiffness equivalent to gain zero as shown in Figs. 8 and 10. During the tests the ground reaction forces were obtained from the strain gauges readings and used to estimate the resultant torques applied to the ankle (Figs. 14 and 15). 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 both at 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

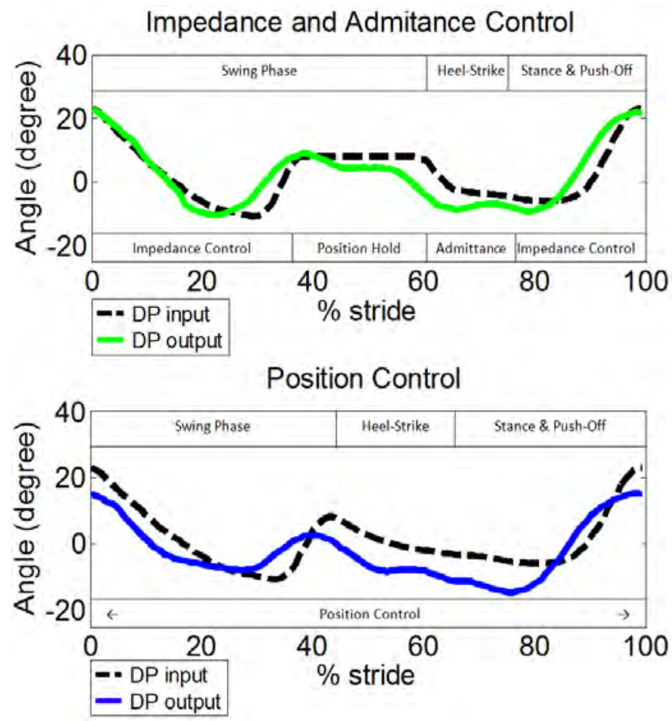


FIGURE 16: Input and output (compensated for 75 milliseconds time delay) of the ankle trajectory in DP during a representative gait cycle with admittance and impedance control (top) and position control (bottom).

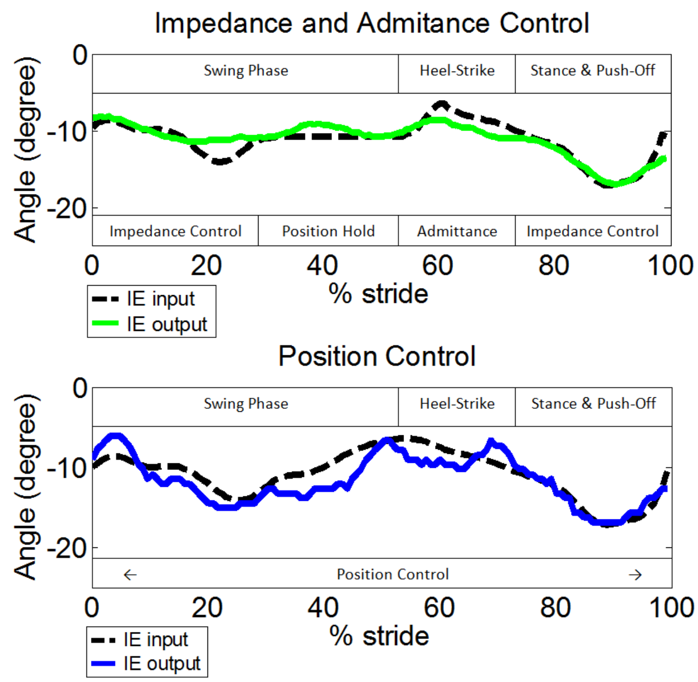


FIGURE 17: Input and output (compensated for 75 milliseconds time delay) of the ankle trajectory in IE during a representative gait cycle with admittance and impedance control (top) and position control (bottom).

Copyright © 2014 by ASME

torques for all experiments, which is expected since the foot is turning left as it walks on the treadmill, putting pressure in the inside edge of the foot.

It is seen in Figs. 14 and 15 that the impedance/admittance controller was capable of reducing the amount of external torque in the foot in both DP and IE; 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 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. The input and output trajectories of the foot in both DP and IE during the tests can be seen in Figs. 16 and 17, 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. From Fig. 16, it can be seen that the impedance/admittance controller input held the ankle constant for near 40% to 65% of the stride, due to the state machine reaching the index of the expected heel-strike, but heel-strike has not happened yet. The impedance/admittance controller was capable of more closely tracking the reference trajectory compared to the position control, since it accounts for the external torques in the control

In IE (Fig. 17), the tracking performance of both controllers decayed compared to the performance in DP. Due to the physical characteristics of the prosthesis, small angular differences between the left and right motors caused larger changes in the foot rotations, making the system more sensitive to disturbances and noise compared to DP.

The tests with the circular treadmill showed that the impedance/admittance controller were capable of better tracking the desired reference trajectory while decreasing the maximum reaction torques in the foot. Future tests are required to measure the specific contributions of the admittance and impedance controller on the performance improvement, and also how the performance could be improved if only one type of controller was used. In both DP and IE directions, the external torques at both heel strike and push off were greatly reduced at the cost of an extended contact time of the foot with the ground. This characteristic needs to be addressed in our future work with a force feedback gain which enables mimicking the time varying human impedance during the stance phase. In IE, an increased external inversion torque was developed due to the constraints imposed by the turning disk, and the impedance control was capable of accommodating and reducing this external torque. The finite state machine worked as predicted and was capable of properly switching to admittance control at heel strike, and switching back to impedance control at push off. Also, the finite state machine was capable of adjusting to the stride duration by advancing the foot to the heel strike angle and holding that position until heel strike was detected.

FUTURE WORK

The torque profile and the performance of the control strategies described will be evaluated at different gait speeds while the prosthesis walks on the treadmill. Also, the effects of changing the radius of turn will be explored. These experiments will give insight on the effects of walking speed and radius of turn on the ground reaction forces and controller performance. The current impedance/admittance controller does not account for IE torques at the heel strike. Future work will evaluate the necessity of estimation of the IE torques at heel strike. Additionally, the current controller does not incorporate the actual time-varying impedance of the human ankle into the control strategy. A proper impedance control dynamically maps the time history of ankle angles into the appropriated time history of ankle torques. The ankle angles can be easily obtained using a motion capture camera system; however, the measurement of the ankle impedance during the gait has yet to be determined. We are currently working on an instrumented walkway platform capable of applying perturbation to the ankle and recording the evoked angle motion during the gait, to estimate the time-varying mechanical impedance of the ankle during the stance period of gait. Consequently, a possible control strategy is to adjust the torque feedback gain of the impedance controller in real time based on pre-recorded look-up data of the ankle impedance trough the stride cycle. Also, different gait scenarios can be studied such as walking uphill, downhill, climbing steps, turning around corners, etc, so the ankle angles and ankle impedance data can be estimated. These different gait scenarios will also be evaluated at different gait speeds. In the prosthesis, a more complex finite state machine will be developed to switch between different gait scenarios and the appropriate lookup data. For the finite state machine feedback, other than the strain gauges, an inertial measurement unit may be used to enhance the controller's efficiency.

CONCLUSION

In this paper, we presented the preliminary steps towards development of a finite state machine for control of a multi-axis ankle-foot prosthesis. The state machine was capable of switching between impedance and admittance control in both Dorsiflexion-Plantarflexion (DP) and Inversion-Eversion (IE). Strain gauges were installed on the prosthesis' foot that were successfully used to estimate the external torques in DP and IE. The estimated torques were used for torque feedback in the impedance and admittance controllers. 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. This showed that the impedance controller was capable of modulating the stiffness of

Copyright © 2014 by ASME

the ankle in a predictable manner. 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 with a fixed torque feedback gain of 0.5 for both DP and IE was capable of better tracking the desired reference trajectory compared to the position control and when there was no control, while decreasing the reaction torques at the foot.

ACKNOWLEDGEMENT

This material is based upon work supported by the National Science Foundation under CAREER Grant No. 1350154.

REFERENCES

[1] Ventura, J. D., Segal, A. D., Klute, G. K., and Neptune, R. R., 2011, "Compensatory Mechanisms of Transtibial Amputees During Circular Turning," Gait & Posture, 34, pp. 307–312.
[2] Colborne, G. R., Naumann, S., Longmuir, P. E., and Berbrayer, D., 1992, "Analysis of Mechanical and Metabolic Factors in the Gait of Congenital Below Knee Amputees," American Journal of Physical Medicine & Rehabilitation, 92, pp. 272 - 278.
[3] Molen, N. H., 1973, "Energy/Speed Relation of Below-Knee Amputees Walking on Motor-Driven Treadmill," Internationale Zeitschrift für angewandte Physiologie einschließlich Arbeitsphysiologie, 31(3), pp. 173-185.
[4] Herr, H. M., and Grabowski, A. M., 2010, "Powered Ankle-Foot Prosthesis Improves Metabolic Demand of Unilateral Transtibial Amputees During Walking," eds.
[5] Ferris, A. E., Aldridge, J. E., Sturdy, J. T., and Wilken, J. M., 2011, "Evaluation of the Biomimetic Properties of a New Powered Ankle-Foot Prosthetic System," eds., Long Beach, CA.
[6] Glaister, B. C., Bernatz, G. C., Klute, G. K., and Orendurff, M. S., 2007, "Video Task Analysis of Turning During Activities of Daily Living," Gait Posture, 25(2), pp. 289-94.
[7] Glaister, B. C., Orendurff, M. S., Schoen, J. A., Bernatz, G. C., and Klute, G. K., 2008, "Ground Reaction Forces and Impulses During a Transient Turning Maneuver," J. Biomechanics, 41(4), pp. 3090-3.
[8] Hogan, N., 1985, "Impedance Control: An Approach to Manipulation: Part I—Theory," J. Dyn. Sys., Meas., Control 107(1), pp. 1-7.
[9] Carignan, C., and Cleary, K., 2000, "Closed-Loop Force Control for Haptic Simulation of Virtual Environments," Haptics-e, 1,pp. 01-1.
[10] Rastgaar, M., Ho, P., Lee, H., Krebs, H. I., and Hogan, N., 2009, "Stochastic Estimation of Multi-Variable Human Ankle Mechanical Impedance," eds., Hollywood, CA.
[11] Rastgaar, M., Ho, P., Lee, H., Krebs, H. I., and Hogan, N., 2010, "Stochastic Estimation of the Multi-Variable Mechanical Impedance of the Human Ankle with Active Muscles," eds., Boston, MA.
[12] Lee, H., Ho, P., Rastgaar, M., Krebs, H. I., and Hogan, N., 2010, "Quantitative Characterization of Steady-State Ankle Impedance with Muscle Activation," eds., Cambridge, MA.
[13] Rouse, E., Hargrove, L., Perreault, E., and Kuiken, T., 2012,
[14] Rouse, E., Hargrove, L., Perreault, E., Peshkin, M., and Kuiken, T., 2013, "Development of a Robotic Platform and Validation of Methods for Estimating Ankle Impedance During the Stance Phase of Walking," Journal of biomechanical engineering.
[15] Lee, H., Ho, P., Rastgaar, M., Krebs, H. I., and Hogan, N., 2013, "Multivariable Static Ankle Mechanical Impedance with Active Muscles," Neural Systems and Rehabilitation Engineering.
[16] Ho, P., Lee, H., Krebs, H. I., and Hogan, N., 2009, "Directional Variation of Active and Passive Ankle Static Impedance," eds., Hollywood, CA.
[17] Goldfarb, M., 2010, "Powered Robotic Legs – Leaping toward the Future," National Institute of Biomediacl Imaging and Bioengineering.
[18] Sup, F., Bohara, A., and Goldfarb, M., 2008, "Design and Control of a Powered Transfemoral Prosthesis," The International Journal of Robotics Research, 27, pp. 263-273.
[19] Sup, F., Varol, H. A., Mitchell, J., Withrow, T. J., and Goldfarb, M., 2009, "Preliminary Evaluations of a Self-Contained Anthropomorphic Transfemoral Prosthesis," IEEE ASME Trans Mechatron, 14(6).
[20] Sup, F., 2009, "A Powered Self-Contained Knee and Ankle Prosthesis for near Normal Gait in Transfemoral Amputees," Ph.D. thesis, Vanderbilt University, Nashville, Tennessee.
[21] Herr, H. M., and Grabowski, A. M., 2012, "Bionic Ankle-Foot Prosthesis Normalizes Walking Gait for Persons with Leg Amputation," Proc Biol Sci, 279(1728), pp. 457-64.
[22] 2012, Bionic Technology with Powered Plantar Flexion, http://www.iwalkpro.com/Prosthetists.html
[23] Kang, H. G., and Dingwell, J. B., 2008, "Separating the Effects of Age and Walking Speed on Gait Variability," Gait Posture, 27(4), pp. 572-527.

Copyright © 2014 by ASME