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Preliminary Design and Evaluation of a Multi-axis Ankle-Foot Prosthesis

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Abstract— **The human gait shows significant differences in the ankle movements during turning and sidestep cutting compared to straight walking, especially in frontal plane. This suggests that the next advancement in lower extremity assistive devices is to extend their design and control to the frontal plane. In this paper, the concept of a multi-axis powered ankle-foot prosthesis is introduced and its feasibility is shown by a proof of concept prototype of a cable-driven, multi-axis ankle-foot prosthesis. The design kinematics and its ankle joint's mechanical impedance in non-load bearing conditions are evaluated and discussed. It is shown that the developed prototype is capable of closely mimicking the ankle movements in both sagittal and frontal planes during turning and walking on straight path with passive mechanical impedance in sagittal and frontal planes comparable to the ones of the human ankle.**

I. Introduction

Recent advances in powered prostheses promise to significantly improve the quality of life and well-being for individuals with impaired mobility. The ankle joint of lower extremity powered prostheses currently commercially available are capable of controlling only one degree of freedom (DOF) in the sagittal plane, focusing on improved mobility in straight walking. Turning, however, plays a major role in daily living activities and requires ankle torques in both sagittal and frontal planes. Studies of four representative daily activities show that turning steps may account for an average of 25% of steps, ranging from 8% to 50% of all steps depending on the activity [1]; which amputees accomplish using different gait strategies than non-amputees. While a non-amputee relies on the hip movement in the frontal plane and moment generated in the ankle joint, an amputee using a passive prosthesis relies on hip extension in the sagittal plane [2-5]. During a turn, modulation of ankle impedance in sagittal and frontal planes plays a major role in controlling lateral and propulsive ground reaction forces in order to accelerate the body center of mass along the gait path; thus, during a step turn, lateral and propulsive impulses are greater compared to a straight step [6]. This difference will result in different gait strategies between amputees and non-amputees to compensate for the lack of propulsion in their passive prostheses to increase maneuverability [2]. This suggests that an ankle-foot prosthesis capable of generating torques in two DOFs, i.e. dorsiflexion-plantarflexion (DP) and inversion-eversion (IE) directions, with impedance modulation similar to the human ankle will augment maneuverability and mobility that leads to a more agile gait. Additionally, design features that allow walking in arbitrary directions on slopes while conforming the foot to the ground profile and uneven surfaces may result in a more efficient gait.

It has been shown that a powered ankle-foot prosthesis reduces the metabolic costs of unilateral transtibial amputees during straight walking by providing sufficient power during push off [7, 8]. 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 [9-12]. 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 [13] developed a microprocessor-controlled artificial foot that limits the increase in metabolic cost to 14% compared to 23% that occurs with a passive prosthesis. The positive work by the prosthesis at the push-off partially compensates for the dissipative negative work at the heel strike of the trailing foot and lowers the redirection of the body's center of mass velocity [14-18]. 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 [19-22]. Also, Hitt et al. designed an ankle-foot prosthesis using a lightweight robotic tendon actuator that provided the majority of peak power for push-off [23, 24]. Au et al. developed an ankle-foot prosthesis [25-27] that later transitioned into a commercially available ankle-foot prosthesis, BiOM.

While the aforementioned prostheses have advanced state-of-the-art design, they are specifically designed for different gait scenarios in the sagittal plane and have one DOF in the sagittal plane. The design strategy for lower extremity prostheses may be improved by incorporating an additional DOF in the frontal plane. The first design for an ankle-foot prosthesis with two active degrees of freedom was presented by Bellman, et al. The design used two identical motors to generate torques in DP and IE [28]. In this paper, we introduced a different design for a powered ankle-foot prosthesis with two controllable DOFs and showed the feasibility of this design by a proof of concept prototype. Further, the design kinematics and its mechanical impedance in non-load bearing conditions were evaluated and discussed.

II. Ankle Rotations and Design Considerations

A series of experiments were performed to quantify the kinematic behavior of the ankle in the context of agility of gait. The experiments used a motion capture camera systems to measure the ankle rotations during stance period of step turn and sidestep cutting for five young subjects with no reported history of biomechanical disorders. In a step turn maneuver, the participants changed the walking direction by

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pivoting around the leading leg and rotating into a new direction approximately perpendicular to the initial direction such as turning around a corner. In a sidestep cutting maneuver, the participant pushed their body sideways using the leading leg while walking forward at near 45 degrees from the original path without rotating the body, similar to stepping sideways to avoid an obstacle on the walking path. The tests were performed at 96 steps per minute and 114 steps per minute, respectively and the results were compared to the results of ankle rotation during straight walking.

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. Remarkably, the push off at straight step and low speed showed a 2.8° eversion, while during the turn, it changed to 10.5° inversion. A more pronounced deviation during the fast speed was observed with 2.5° eversion and 14.3° inversion for straight step and step turn, respectively. Those values were equivalent of 248% and 312% of the straight step IE average range of motion (ROM) 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. A significantly different behavior is seen in the 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 was 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 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.

Both experiments implied that in order to have an agile ankle-foot prosthesis, design considerations should include a controllable DOF in the IE direction.

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 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 in a 2-DOFs ankle (Fig. 1). 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 a socket that will be tailored to fit to the residual limb of the amputees (not shown) and connects to the pylon (A). Two DC motors (E) and planetary gear heads (D) are powered by two motor controllers (B) and receive signals from a DAQ board (M) connected to a remote computer and two quadrature encoders (I). Two cable drums (J) transfer the required torque to the ankle through the shock-absorbing nylon rope (K). The cable is attached to the cable drum and slippage is avoided by looping the cable around two bolts, when tightened, provide enough friction to transfer the forces to the cable. The cable drums are designed to rotate up to 270° to avoid the cable from releasing itself from the drum. A universal joint (F) 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 (C). The cable is attached to a carbon fiber plate (H) which is connected to a commercially available prosthetic foot (Otto Bock Axtion®) (L). 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 (G). 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.

The proposed mechanical design with two DOFs uses the fact that three points are sufficient to define a plane in space. If the plane is constrained from translations and has a fixed point of rotation, by defining the position of the three points, all the rotations of the plane can be controlled. A simplified schematics of this concept is shown in Fig. 2. It is seen that three intersection points A, B, and C can be moved by the cables, and thus generate the rotations about the X and Y axes which are equivalent to DP and IE, respectively. If the two motors' driving forces are in the same direction, for example in the negative Y direction, they cause the points A and B to move downwards, while point C moves upwards, resulting in dorsiflexion. If the motors move in opposite directions, for example if the right motor applies a force in the positive Y direction and the left motor applies a force in the negative Y direction, point A moves upwards, while point B moves downwards, generating an eversion torque in the foot; and since point C is located on the axis of symmetry of the plate along the X direction, and the cable goes through a pulley at that point, it neither moves in any direction nor constrains the cable from motion.

Providing sufficient power and torque for the ankle joint without significant increase in the weight of the powered prosthesis is a challenging issue. Dual motor systems have been previously used to produce the appropriate torques in only DP direction [23]. The proposed design in this paper however benefits from using two motors to produce torques in both DP and IE directions, simultaneously. Considering the fact that more power is required in DP than IE during walking, the design benefits from two identical motors instead of having a larger motor for DP and one smaller motor for IE. Also, it does not require any extra hardware other than the universal joint at the ankle when compared to one DOF dual motor designs.

The cable driven design, besides the ability to control the ankle in two DOFs, may provide a significant flexibility in managing the inertia of the ankle-foot prosthesis too. The motors and gear boxes are not constrained to the ankle, meaning they can be relocated to accommodate the characteristics and preferences of the user. The system has the potential to move up or down on the pylon to accommodate to the users' residual limb or to optimize the weight distribution.

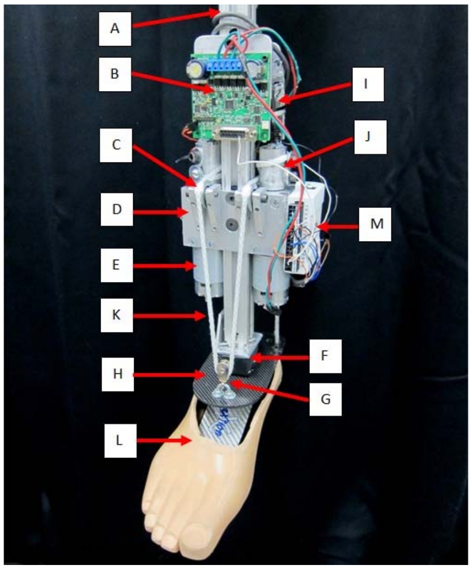

Figure 1. Prototype of the powered ankle-foot prosthesis with two DOFs.

IV. Design Analysis

The energy consumption required at each step in an average able-bodied human weighing 80 kg is 36 J for walking (250 watts peak power)[24, 29] with 140 Nm maximum torque [30]. These amounts are 35% higher for an individual with a transtibial prosthesis [24, 31] and the prosthesis is estimated to have up to 40% losses; resulting in an anticipated peak power consumption of 470 watts and an energy consumption of 68 J during each step. Considering those requirements, the prototype prosthesis uses two brushed DC motors capable of a continuous torque output of 0.25 Nm at 9200 RPM (producing 240 watts each for a total of 480 watts); each connected to a gear box with a 104:1 reduction gear box. Two 11.1 Volts LiPo batteries connected in series with an energy density of 572 kNm/kg are estimated to provide energy for 5800 steps are used. The current prototype weighs 3 kg without the batteries and socket which is the equivalent weight of the ankle-foot of a 90 kg person [30].

The motors and gear boxes are capable of applying 2 kN of continuous pulling force in the cable (ignoring losses) at 1.17 m/s. The moment arm in DP (the distance from the center of the universal joint to the cables in the back of the carbon fiber plate located on the longitudinal axis of the foot) is 57 mm, and the moment arm in IE (the distance from the center of the universal joint to the cables in the back of plate along with the lateral axis of the foot) is 28.4 mm. Since both motors contribute to DP rotation, the maximum theoretical

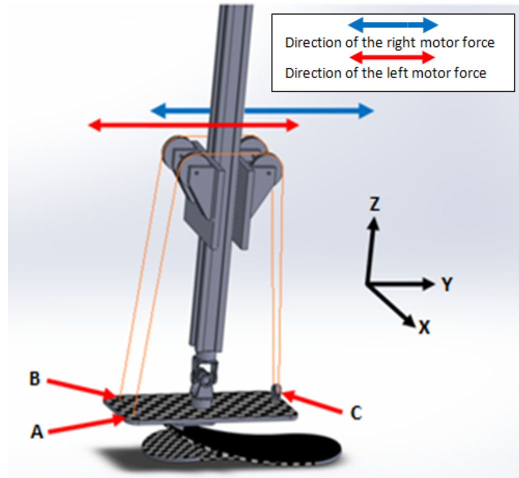

Figure 2. Simplified drawing of the cable driven mechanism showing intersection points A, B, and C between the cable and the carbon fiber plate. For simplicity, the motors and cable drums are not shown.

continuous torque in DP is 228 Nm at 20.5 rad/sec. Maximum IE torque is 57 Nm at 41 rad/sec. Due to the geometry, the effective moment arms change as the foot rotates from the central position resulting in a 7% loss in DP torque at -21.6° of plantarflexion (maximum DP angle [32]) and a 6% loss in IE at 19.7° of inversion (maximum IE angle [32]). The torque values and angular velocities are still within the target range even with 50% losses.

For the cable, a nylon rope was chosen because of the strength, flexibility, and capability of providing shock absorption. The carbon fiber plate is 3.175 mm thick and is a fundamental component of the design acting as a spring connected in series between the cable and the foot. The cable needs to always be in tension to assure the proper control over the foot, causing the carbon fiber plate to always be under a bending moment.

V. Evaluation of the Design Concept

The developed prototype was evaluated to meet two criteria. First, the design kinematics should be capable of reproducing the same ankle rotation as the human ankle during the stance and swing periods of gait. Second, the impedance of the prototype's ankle in dorsiflexion-plantarflexion and inversion-eversion needed to be comparable to the mechanical impedance of the human ankle, where the mechanical impedance of the ankle maps the time-history of angular displacements onto the corresponding time-history of torques at the ankle joint.

A. Kinematic Evaluation

Currently, two optical quadrature encoders (200 pulses per revolution) provide position feedback to a remote computer that uses a proportional plus derivative 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 left ankle during the sidestep cutting 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. 3. 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 test the robot was moving at 50% of the walking speed. For ease of comparison, the output plot has a time shift to remove the 24 milliseconds respective delay of the output. The current prototype was developed as a proof of concept to validate the design kinematics and uses low-cost brushed DC motors and gear boxes with limited bandwidth; which are the cause of the observed time delay. Therefore, faster motors and sensors with lower noise levels will be used in future designs. Both plots indicate the mechanism is capable of reproducing the same ankle rotations as the human subjects during step turn and sidestep cutting.

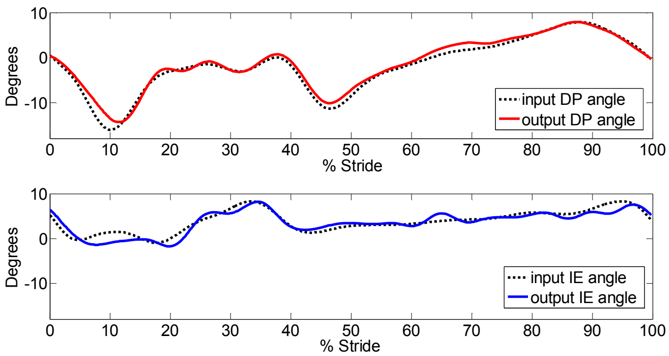

Figure 3. Input and compensated output for time delay (24 milliseconds) of the ankle-foot prosthesis. The input is the recorded right ankle rotations of a representative subject during swing and stance periods of the sidestep cutting at fast speed.

B. Mechanical Impedance Estimation

Recently, hierarchical control strategies have been developed for impedance control of active prostheses [20, 33]. The higher level control identifies the gait cycle, and the lower level control regulates the actuators for proper impedance characteristics. Following the same strategies, an ankle-foot prosthesis can be designed to have an initial and passive mechanical impedance similar to a human's ankle in a stationary position. This objective requires a proper design of the ankle joint components. This feature may provide faster modulation of the ankle impedance based on the state of the gait cycle.

The purpose of this evaluation was to identify the passive impedance of the prosthetic device ankle and compare it to the impedance of the human ankle. The impedance estimation experiment setup was similar to the procedure reported in [34, 35], where an Anklebot®, a multi-axis lower extremity therapeutic robot, was used to apply pseudo-random torque perturbations to the ankle joint in DP and IE directions, simultaneously. The applied perturbations had a bandwidth of 100 Hz and the provoked ankle angles were record. A multivariable stochastic system identification method was used to estimate the mechanical impedance of the ankle using the collected data. Similar procedures were used for estimation of passive mechanical impedance of the ankle-foot prosthesis while all its controllers were turned off and no external load was present other than the torques provided by the Anklebot. The experimental setup can be seen in Fig. 4, where the two devices were attached mechanically to each other. The Otto Bock Axtion® foot and its rubber foot shell were inserted in the same type of shoe used in human tests to ensure consistency in the experiments. Similarly, the same test was done on a representative human subject for comparison with the prosthesis. Impedance test of the human subject was performed with no external load and in both relaxed muscles and 10% of their maximum voluntary contraction (MVC) of the tibialis anterior. The muscle contraction was monitored using surface electromyography (EMG) following the procedures described in [34]. The EMG signals were monitored using a Delsys Trigno Wireless EMG System® with surface electrodes placed at the belly of the tibialis anterior. The EMG signal was sampled at 2 kHz and the root-mean-square value of a window of 13.5ms of data calculated and displayed on a computer screen in order to provide a visual feedback to the participant for maintaining constant muscle activity. The results can be seen in Fig. 5 and 6.

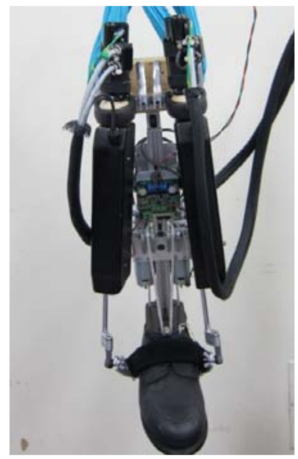

Figure 4. Anklebot attached to the prosthetic foot for estimation of the mechanical impedance of the ankle-foot prosthesis.

Fig. 5 shows the Bode plots of the mechanical impedances in the DP direction of the prototype prosthesis, the human subject's ankle with relaxed muscles, and the human subject's ankle with 10% MVC. The quasi-static stiffness of the prototype prosthesis, which is the impedance magnitude at low frequencies, was 39.5 dB (94 Nm/radian) in DP at 1 HZ. Also, it showed a relatively linear impedance and phase over the frequency range of interest (0 to 5 HZ), with an average coherence value of 0.92. The human subject showed similar stiffness in DP for the co-contraction test and lower stiffness in the passive test when compared to the prosthesis. IE impedance magnitude (Fig. 6) of the prosthesis was between the active and passive stiffness of the human sample with a value of 24.24 dB (16 Nm/radian) at 1 Hz. Considering that the magnitude of the ankle impedance is different among the individuals, the results showed that it was feasible to design the mechanism with the impedance characteristics close to the human ankle impedance.

VI. Discussion

In an ankle-foot prosthesis, a mechanical impedance characteristic similar to the human ankle along with the capability of mimicking the kinematics of the human ankle during different gait maneuvers will increase the agility of the gait in amputees. The control of the ankle joint in the ankle-foot prostheses should be similar to the time-varying impedance of the ankle in two DOFs during different phases of the stance period while providing the required torque. The implemented mechanical design, although in early stages of development, showed anthropomorphic characteristics. The design was successful at mimicking human ankle motion, and

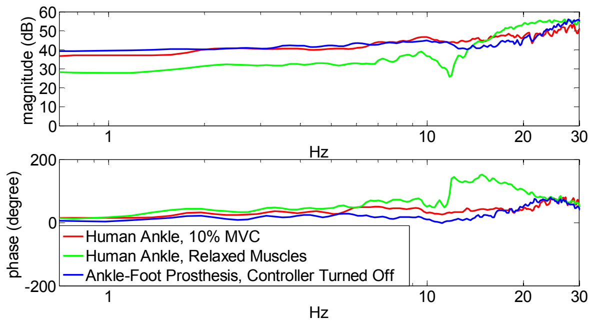

Figure 5. Plots of the magnitude and phase of the ankle impedance in DP for the prosthetic robot and a human subject with relaxed muscles and 10% MVC.

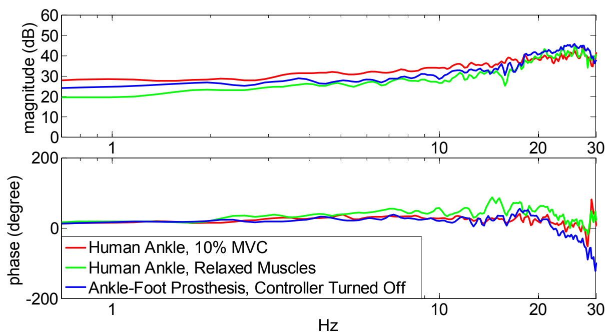

Figure 6. Plots of the magnitude and phase of the ankle impedance in DP for the prosthetic robot and a human subject with relaxed muscles and 10% MVC.

showed comparable mechanical characteristics during step turn and sidestep cutting. The mechanical impedance can be improved through design development and implementing appropriate impedance controllers in the future.

While the maximum lifting force of the robot was generated, the carbon fiber plate flexed due to the applied torque. Since the encoders read the cable displacement instead of foot angles, the controller perceived the deflection of the carbon fiber plate as an angular displacement of the foot. This caused the position controller to reduce the torque being applied prematurely, and thus the maximum lifting force was less than anticipated, although it was capable of lifting a 72 kg person. Future designs will employ strain gages to measure the composite plate flexion to increase the precision of the position and the resulting torque. Also, rotary encoders at the foot may provide the actual foot angles. Additionally, the current DC motors are not sufficiently fast, even though their torque and powers meet the design criteria. The motors will be upgraded to improve the design in the future.

Testing the ROM in IE, it was found that the IE motion becomes unstable at angles above 62° (if an external moment is applied). This is the equivalent of rolling the ankle, which is a common injury among active people and are mostly in inversion [36]. In the prototype prosthesis, ankle rolling happens when either points A or B (Fig. 2) crosses the Y axis. At this point, the tension in the cable, which is applying a torque against the disturbance force, changes the direction of the torque and makes the ankle unstable, however it is below the ankle's ROM during different gait maneuvers reported here.

VII. Conclusion

It was shown that the rotation of the ankle in inversion-eversion significantly changed during turning 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 controllable degrees of freedom in sagittal and frontal planes. Evaluation experiments showed that the mechanism is capable of reproducing the human ankle rotations during step turn and side step cutting. Additionally, the ankle joint was designed with mechanical impedance characteristics similar to the human ankle with 10% maximum voluntary contraction in both sagittal and frontal planes, suggesting the feasibility of the design in mimicking the dynamics of the human ankle.

VIII. Acknowledgement

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