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Amputee Subject Testing Protocol, Results, and Analysis of a Powered Transtibial Prosthetic Device

A powered ankle-foot prosthesis and its control system were previously designed and built. To evaluate this prosthesis, amputee subject testing was performed. The testing results are analyzed and compared between the powered prosthesis, passive prosthesis, and able-bodied gait. Qualitative comparison showed the prosthesis achieved the design objectives. During stance phase, active ankle moment was generated in the powered prosthesis before push-off to help the amputee walk more naturally. During swing phase, the powered prosthesis was able to move to natural position to achieve foot clearance. However, the prosthesis is slightly under powered compared with the able-bodied ankle. [DOI: 10.1115/1.4027497]

1 Introduction

Most of the current commercially available transtibial prostheses utilize a passive foot/ankle system. They essentially function as a spring in combination with a damper that absorbs energy during the early stage of gait and releases the stored energy at the end of the stance phase before push-off. However, studies show that the muscles surrounding an able-bodied ankle joint actively produce energy to achieve natural push-off [1]. This function does not exist in current passive transtibial prostheses. Because active power is required for adequate push-off, a new type of powered transtibial prosthesis with a brushed DC motor was previously designed and fabricated [2,3]. To regulate the power generated by the motor, a control system was also designed and implemented [4]. Amputee subject testing is necessary to provide experimental data in order to verify the design concept of this prosthesis.

Multiple academic groups have designed and built powered transtibial prostheses and have performed amputee subject testing of their devices. "BioM," the transtibial prosthesis developed at MIT, was tested on amputee subjects where metabolic cost was used as the major criteria for the performance. It was shown that the designed powered prosthesis had better performance than conventional prostheses [5–7]. The latest developed neuromuscular control system was also evaluated on amputee subjects where the results showed this new control system was able to adapt to inclined ground and had comparable performance to a biological ankle [8]. The transtibial prosthesis "SPARKY" designed at Arizona State University was tested over several months on a specially built treadmill. The data shows the device provides ankle joint motion and ankle power that is comparable to able-bodied persons [9,10]. A transfemoral prosthesis developed at Vanderbilt University has both powered knee and ankle joints. It was successfully tested on amputee subjects with the subjects being able to walk on ascending and descending slopes [11–13].

The design concept in this research is different from the approach of the other groups. The nonlinear nature of a four-bar

mechanism with respect to its angular position was utilized in conjunction with a torsional spring to mimic the nonlinear behavior of a human ankle [3]. Simulation shows that the mechanism matches the nonlinear impedance behavior of a human ankle [2]. The advantage of this design is that it can give the user a smoother transition between the nonlinear points of ankle impedance. In addition, this design is mechanically simpler than other designs because only three major components exist in this design: The four-bar mechanism, torsional spring, and brushed DC motor.

Amputee testing was performed to evaluate the prosthesis and its control system. Preliminary success was achieved in that the designed prosthesis functioned consistently with the control objectives. The structure of this article is as follows. In Sec. 2, the previous work on the design and implementation of the mechanism and its control system is briefly described in Ref. [3]. The amputee subject testing protocol is then explained. The results of the amputee subject testing are illustrated and analyzed in Secs. 3 and 4. The conclusion is made in Sec. 5.

2 Methods

The amputee subject testing in this research is for a previously designed powered ankle-foot prosthesis [3]. The prosthetic mechanism design uses a four-bar mechanism in combination with a torsional spring to closely match the nonlinear impedance behavior of a human ankle with respect to angular position. A control system was designed and implemented in order to boost the moment generated by the mechanism to match an able-bodied person's ankle moment profile with respect to time.

2.1 Previous Work. A sketch of the mechanical design of the prosthesis is shown in Fig. 1. The active power from the motor is acting at joint C, which is the same position as the torsional spring. The length of the links was optimized to create a nonlinear moment profile with respect to ankle angular position during the stance phase.¹ The optimization was completed based on the

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¹Stance phase is defined as the period when the foot has contact with the ground. In contrast, swing phase is defined as the period when the foot has no contact with the ground and thus in swing.

Fig. 1 The mechanism configuration of the designed prosthesis

Fig. 2 Prosthetic device prototype

principle of minimizing the square errors between the theoretical moment and optimized moment plus an energy penalty function. Ankle moment data from Winter [14] was used as the moment benchmark. Only normal speed of able-bodied walking was considered in this optimization to prove the feasibility of the design

concept. The details of this optimization are described in Bergelin et al. [2].

After going through the standard mechanism design process, a prototype was fabricated (Fig. 2) and passively bench tested to verify its operation. Even though slightly underpowered, the four-bar mechanism and torsional spring were able to provide the correct profile shape close to the theoretical moment response during stance phase [3,4]. In order to reproduce the full function of an able-bodied person's ankle, a control system is required to regulate the additional power provided by the motor.

After studying the dynamics of gait and the constraints of the mechanical design, it was determined there should be two objectives for the control system. First, the motor needs to generate extra energy to help with the push-off at the end of the stance phase. Second, the motor needs to move the foot plate of the prosthesis from plantar flexion to slight dorsiflexion position to have foot clearance during the swing phase. A two-level finite state control (FSC) algorithm was determined to be best fit for these control objectives and is illustrated in Fig. 3.

To realize the designed control algorithm, a digital signal processor (DSP) was selected to be the processing unit. The control algorithm was programmed using MATLAB SIMULINK and converted into executable and downloaded to the DSP. A Maxon EPOS 50/5 motor controller was selected as the lower level motor drive. Sensors were selected to provide the required feedback to the controller. The details of the control system design were explained in Sun and Voglewede [15]. With the mechanism and control system designed and built, amputee subject testing needs to be performed to verify the design theory and provide experimental results for performance evaluation and further improvement.

2.2 Amputee Subject Testing Protocol. In order to reduce complexity, only level walking was tested. The testing was performed by an amputee walking at a self-selected walking speed within a Vicon motion capture laboratory at the Medical College of Wisconsin Department of Orthopaedic Surgery's Center for Motion Analysis. The prosthesis was placed on the left leg of a healthy unilateral below-knee amputee. Compared to the amputee subject's passive prosthesis, the tested powered prosthesis has extra components including controllers and batteries mounted on a backpack carried by the subject. The powered prosthesis was connected to the pylon by using standard socket connection.

The amputee subject chosen is a traumatic amputee with K4 functional activity level and a mass of 86.5 kg. The passive prosthesis he uses is endoskeletal prosthesis, which he has been using for eight years. No other medical conditions affect his gait. An institutional review board approved consent form was signed by

Fig. 3 Control algorithm to regulate the powered generated by the motor

the amputee subject following full explanation of the testing procedure. Reflective markers were placed on the amputee subject via a modified Helen-Hayes marker set to obtain testing data with Vicon MX cameras capturing the movement of the markers. There are 12 Vicon MX40 cameras working at 120 Hz. The markers were placed on what was the ankle axis of the prosthesis as well as the shaft for the “calf” wand. There was also one marker on the heel and one at the same height of the heel marker in the area representing the base of the toes. The marker placement did change between passive prosthesis and powered prosthesis because the shape of the two prostheses feet was different. The other markers were consistent throughout the testing. First, the subject walked using his own passive prosthesis. Then, the subject was allowed time to walk on the powered prosthesis to acclimate to it. The powered prosthesis testing began after the subject felt comfortable with it.

The subject was asked to walk along a 10 m walkway at his most comfortable speed. Force plates embedded along the walkway measured the ground reaction forces (GRF). The force plates were two Bertec 6-DOF and two AMTI OR-6 6-DOF plates which collect data at 3000 Hz. In order to reduce any changes in gait, the amputee subject was instructed to ignore the existence of the force plates. The kinematics, power consumption, the moments at each joint, which includes ankles, knees, and hips on both sides, and other dynamic results were calculated using the standard inverse dynamics model (Vicon Plug-In Gait Model) in Vicon’s NEXUS 7.1 software. The standard filter for the Plug-In Gait model, Woltring filter with an mean square error value of 20, was used. Even though the difference in the masses between the prostheses and the natural leg can affect the kinetic calculations, anthropometric and kinetic equations were not modified in the Vicon Plug-In Gait Model. Therefore, the kinetic results from the amputee subject testing only indicate the performance of the powered prosthesis qualitatively.

To obtain satisfactory kinetic results, the subject must fully step on the foot plate to enable complete GRF information to be collected and the kinetic results to be calculated using the inverse dynamics model. A total of 15 trials were performed with the amputee subject wearing their own passive prosthesis. In 13 of these trials, the subject did not fully step on the force plate; therefore, no complete GRF information was collected from the foot plate. Satisfactory kinetic data were only obtained for two trials. Kinetic data including both the natural leg and passive prosthesis were used for data processing. For the powered prosthesis testing, 27 trials were performed, and for three trials, the subject fully stepped on the force plate. Kinetic data including both the natural leg and powered prosthesis for those three successful trials were used for data processing. Therefore, there are five successful trials for the natural leg for both testings, two successful trials for the passive prosthesis and three for the powered prosthesis. The data from those successful trials were used for results comparison.

2.3 Data Processing. The testing results are interpreted from three perspectives. First, the results are compared between each trial the amputee subject completed while wearing the powered prosthesis. The purpose of this comparison is to verify the consistency of the performance of the powered prosthesis. Second, the results are compared between the powered prosthesis, the natural leg, and the benchmark data from Winter [14]. The purpose of this comparison is to discuss if the performance of the powered prosthesis meets the design requirement and how its performance is compared to the natural leg. Third, the results are compared between the powered prosthesis and the subject’s own passive prosthesis. The purpose of this comparison is to qualitatively verify the performance of the powered prosthesis compared with the passive prosthesis from kinematics and generated moment perspectives. The potential of the powered prosthesis to perform better is investigated.

In order to make the comparison consistent in time, all the data were normalized to 0–100% of gait cycle. The beginning and the

end of one gait cycle were defined as two consecutive heel strikes of the same foot. One critical event needs to be clarified is “push-off,” which is defined as the time instant when the stance leg leaves the ground, which is approximately 62% of gait cycle [16]. For each group, the results are evaluated according to three sets of data: the angular position, the moment, and the power absorption or generation in the ankle. The angular position of the ankle evaluates the kinematic performance. The moment in the ankle evaluates the kinetic performance. The power absorption or generation is presented to support the verification of kinetic performance. The GRF is not critical design criteria for the design. The experimental results of GRF can be found in Ref. [17].

3 Results

3.1 Result Between Trials Wearing Powered Prosthesis.

The ankle angle versus time of the subject wearing the powered prosthesis for three trials is shown in Fig. 4. It can be seen that the kinematic performance of the prosthetic ankle is consistent. The only difference that is noticeable is that, for trial 3, during the loading response phase, the prosthetic ankle is 4 deg less plantar flexed than the other two trials.

The prosthetic ankle moment versus time comparison is shown in Fig. 5. The peak moment of these three trials are the same. The difference lies in that the moments in trial 2 start to accumulate in the prosthesis earlier than trial 1 and trial 3; the moments in trial 1 releases later than trial 2 and trial 3.

Fig. 4 The ankle angle comparison between the three trials of the powered prosthesis

Fig. 5 The ankle moment comparison between the three trials of the powered prosthesis

Fig. 6 The ankle power comparison between the three trials of the powered prosthesis

Fig. 8 The ankle moment comparison between the powered prosthesis, the natural leg and Winter's data

Fig. 7 The ankle angle comparison between the powered prosthesis, the natural leg and Winter's data

The power absorption or generation results are shown in Fig. 6. The negative y axis means the prosthesis ankle is absorbing energy and the positive y axis means prosthesis generating energy. The peak power absorption and generation is 1.02 and 2.29 W/kg for trial 1, 1.31, and 2.76 W/kg for trial 2, and 0.89 and 1.75 W/kg for trial 3, respectively.

3.2 Result Between the Prosthesis, Natural Leg, and Winter's Data. The ankle angle data of the powered prosthesis testing, natural leg, and Winter's data [14] are shown in Fig. 7. It can be seen that the powered prosthesis performs more dorsiflexion during the stance phase and performs more plantar flexion during the swing phase. The data between the amputee's natural leg and Winter's data are also different, with the amputee's natural leg performing more dorsiflexion during the entire cycle of the gait.

The ankle moment comparison between the powered prosthesis testing, natural leg, and Winter's data is shown in Fig. 8. Comparing with the amputee's natural leg, the peak moment of the powered prosthesis is 5% smaller. Also, it can be seen that the push-off of the powered prosthesis is 10% of one gait cycle earlier than the natural leg. The power absorption and generation comparison is shown in Fig. 9. It shows that the powered prosthesis absorbs more energy than the other two during the stance phase. The power generation is not as large as the other two during the push-off.

3.3 Result Between the Powered Prosthesis and the Passive Prosthesis. The ankle angle comparison between the powered prosthesis and the passive prosthesis is shown in Fig. 10.

Fig. 9 The ankle power comparison between the powered prosthesis, the natural leg and the Winter's data

Fig. 10 The ankle angle comparison between the powered prosthesis and the passive prosthesis

It is similar to Fig. 7; the powered prosthesis performs more dorsiflexion during the stance phase and more plantar flexion at the end of the swing phase.

The ankle moment comparison is shown in Fig. 11. It shows that the peak moment before the push-off is around 10% larger in the powered prosthesis than the passive prosthesis. Similar to Fig. 8, the push-off happens earlier in the powered prosthesis than the passive one. Figure 12 shows the power absorption and

Fig. 11 The ankle moment comparison between the powered prosthesis and the passive prosthesis

Fig. 12 The ankle power comparison between the powered prosthesis and the passive prosthesis

generation comparison. It can be noticed that the power absorption in the powered prosthesis during stance phase is more than the passive prosthesis. The power generation during the push-off is the same. Therefore, the powered prosthesis generates a stronger push-off.

4 Discussion

The designed prosthetic mechanism consists of a torsional spring connected to the prosthetic ankle joint. The moment stored in the torsional spring during stance phase is proportional to the angular displacement contributed by the amputee subject [4]. Figure 4 shows that the prosthetic ankle performs consistently kinematically in the three trials. This indicates that the moment stored in the torsional spring contributed by the angular displacement of the prosthetic ankle is consistent. The total prosthetic ankle moment is from two sources: the angular displacement of the torsional spring and the motor. The moment provided by the torsional spring is consistent between each trial; therefore, it can be concluded that the moment contributed by the motor is also consistent. These consistencies indicate that the prosthesis and its control system perform consistently between each trial. Therefore, the average of these three sets of data can be compared with the natural leg and the benchmark to see if the performance meets the design requirements.

One major objective of the design of this powered prosthesis is to meet the reference ankle moment of Winter's benchmark, and thus an able-bodied person. It can be seen in Fig. 8 that this goal is partially achieved, with the moment in the powered prosthesis ankle more than that in the passive prosthesis even though lower

than the natural leg. The reason why this objective is not fully achieved is that friction loss is not considered during the mechanism or control system design because of the difficulty in calculating the losses. There are three possible ways to address this problem. The first is to use a dynamic friction model to calculate how much power was lost and overpower the motor. The second way is to use more efficient components or redesign the mechanism to achieve better efficiency. For example, a brushless DC motor or a servo motor could be selected to raise the efficiency or better lubrication could be used. The third way is to use another lower level control algorithm to replace the current proportional-integral-derivative (PID) controllers. As the ankle dynamics of the human gait is nonlinear with respect to the angular position, if more gait patterns are programmed into the higher level FSC in the future, one single set of PID parameters probably will not satisfy the need. Therefore, a more advanced control algorithm could be used for the lower level control, such as nonlinear controller, adaptive control, and some other control algorithm, which could make the generated moment profile match the reference more closely.

The other objective of the design is that the powered prosthesis should go back to the neutral position during swing phase to get ready for the next heel strike. However, as shown in Fig. 7, the powered prosthesis performs around 9 deg more of plantar flexion at the end of the swing phase. This was also corroborated by the amputee subject and the prosthetist during the testing. This may be due to the physical limitation of the torsional spring and the mechanism. During the swing phase position control, the ankle joint was programmed to go to its maximum dorsiflexion position. Going further will break the torsional spring or affect the motor. One possible solution will be to revisit the optimization of the four-bar mechanism to enable the mechanism move to neutral position easier. Another solution is to utilize a more powerful motor. Nevertheless, the amputee subject did state in the interview after the testing that he had enough foot clearance during the testing for the heel strike.

While the push-off happens earlier in the powered prosthesis than in the natural leg as shown in Fig. 8, it is consistent with Winter's data. During the control system design process, Winter's data were used as benchmark in the design; therefore, the timing of the powered prosthesis testing and Winter's data are consistent. Another possible reason for the early push-off is the amputee subject's walking pattern inherently performs a earlier push-off at their amputated side with the passive transtibial prosthesis [18]. In other words, even though the prosthesis was providing active power to assist the push-off, the testing subject may still perform an earlier push-off because they are not familiar to the new system. This difference can be alleviated if the amputee subject were to use the powered prosthesis for a longer period of time.

One advantage the powered prosthesis gives is that it generates 10% more moment than the passive prosthesis during the stance phase, which will help with the amputee's push-off. This is also corroborated by the amputee subject in the interview after the testing that they did feel a push-off from the powered prosthesis when their prosthetic foot left the ground. This indicates that the powered prosthesis achieves the goal of generating more energy in the ankle to help the amputee push-off.

Another advantage of the powered prosthesis is that the power can be adjusted in the control system. As noted in Ref. [17], there is energy loss inside the mechanism; the energy loss was not compensated in the control system in this first preliminary testing. If the power provided by the motor is magnified by a certain gain greater than one, the powered prosthesis is able to provide more moment to help with the push-off.

Kinematics and kinetics data from Winter [14] were used as benchmark in this design. This data may not be the ideal set to use due to inconsistencies in the process and model used by Winter in the past and the modern processes and models used today. However, Winter's data are the most readily available data, which consist both kinematics and kinetics to the best of the authors'

knowledge. The possibility of using other more modern data should be investigated.

5 Conclusion

Based on the amputee subject testing results, it can be concluded that the powered prosthesis performed consistently with the design objectives and achieved better performance than the subject's own passive prosthesis. The active components generated additional power to assist with the push-off. The control system moved the foot plate to neutral position as much as possible to obtain ground clearance. However, this prosthesis still slightly underperformed compared with able-bodied human ankle from two perspectives. First, even though the powered prosthesis achieved better ankle moment than passive prosthesis, it did not achieve the same ankle moment as an able-bodied ankle. Second, during swing phase, the foot of the prosthesis was not able to fully move back to neutral position, which could cause foot-ground interference.

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