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A Novel Design Approach and Operational Strategy for an Active Ankle-Foot Prosthesis

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1 Introduction

In order to enable lower limb amputees to regain natural and versatile walking patterns, the development of active ankle-foot prostheses is required. During locomotion, above-knee amputees show a lower walking speed for keeping the energy effort in the range of non-amputee walking [2]. To avoid this, prostheses with adaptable characteristics and the possibility to store and reuse energy can provide advantages like more dynamic locomotion and secure operation on uneven terrains without increasing the user's energy effort. Using an actuated prosthesis, a supporting torque can be applied to the ankle joint and thus enable users to walk with less energy effort. By this, such components might also protect amputees against medical consequences due to compensation movements [1].

2 State of Art

The large majority of commercially available foot prostheses are passive prostheses with a fixed angle between shank and foot [3]. The applied elastic elements like carbon springs are designed for a basic adjustment to uneven terrain and a partial recuperation of energy during a gait cycle. Further, the mechanical characteristics can be customized by design, e.g. by splitting those springs to support eversion or inversion motions. Semi active foot prostheses do not support users with direct actuation, but are able to adjust position or characteristics of the foot and thus can reduce energy effort and provide more natural gait patterns. Currently, there are two active ankle foot prostheses coming into the market: The iWalk BiOM [4] and the Springactive Odyssey [5]. Further, active prosthetic ankle joints are developed in academic research projects such as the AMP Foot 2.0 [6]. Those prostheses support locomotion with a motor spring complex. While most active approaches use a serial elastic actuator (SEA) composed of a motor, gears to transform rotational into translational motion, and springs, parallel setups of spring and actuator can also be applied.

3 Concept

In [1] it is shown that the creation of a natural walking pattern requires variable ankle stiffness, energy storage as well as active support synchronized with individual gait. The authors' concept to achieve these goals is given in Figure 1.

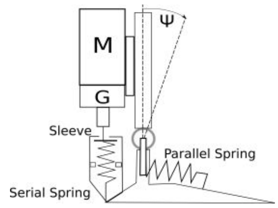

Figure 1: Mechanical setup of the ankle-foot

The drive train consists of two DC motors with a power of 90W and a continuous torque of 0.2Nm. Two ball screws with gear ratios of 6283 are attached to the motors to transform the rotational motion of the motor's shaft into a translational motion of the screws. The lower ends of the screws are connected to two parallel extension springs placed in a sleeve. Each spring can be loaded with 739N and has a maximal extension of 18mm. In the sleeve two mechanical stops limit the maximal extension of the springs. With the lower mechanical stop, the actuator and gears are able to bring a force on the heel without compressing the extension springs, which might result in a damage of those. The upper mechanical stop protects the springs from increased tensile loads. Parallel to the SEA the so-called parallel spring is installed. The spring is connected to the shank on the one side and to the foot on the other side. During dorsiflexion, energy is stored by elastic deformation of the spring due to the angle deviation between shank and foot. Beyond the angle range from -5° to +10°, the spring is inactive due to the kinematics design of the foot. For this, a conventional low-profile carbon device with a split toe is proposed. The split toe enables the device to fulfill an inversion or eversion motion and to adapt on uneven surfaces.

4 Operational Strategy

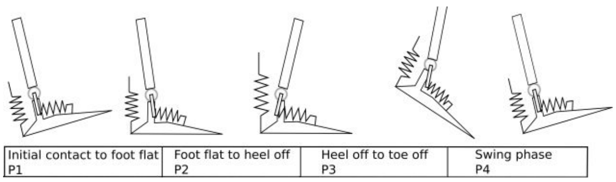

Figure 2: The operational strategy

Figure 2 shows the prosthetic foot in different moments of the gait cycle. In P1 a negative torque is created around the ankle joint axis by the ground reaction forces. The torque will rotate the foot from initial contact to foot flat. During P2 the sign of the external torque switches from negative to positive because the body is moving in walking direction over the on the standing foot. In P2, the parallel spring extends due to the angle deviation between foot and shank and hereby creates a negative torque around the joint axis. The serial spring is extended through the motors and creates a negative torque. At the end of P2, the torques created by the springs are equal to the external torque and the bodyweight is held due to the extensions of both springs. Hence, parallel and serial spring store energy in form of elastic deformation energy. P3 follows with a positive external torque around the ankle joint. In order to support locomotion, the torque created by the parallel and serial spring must be greater than the external torque. This is accomplished by a further tensile load of the serial spring introduced by the motor. The sum of all torques -the external torque and the torques created by the springs- is negative so that the heel starts to rise from the floor. In this phase the parallel spring is deactivated by a sliding mechanism as soon as a compression load acts on it. In the swing phase no external torque appears due to the missing contact to the ground. In this phase, the mechanical stops in the sleeve are used to provide forces on the heel and bypass the serial spring. Hereby, a positive torque around the joint axis rotates the foot to a right angle relative to the shank.

5 Results and Discussion

To survey the design concept and operational strategy, simulations of the system dynamics are conducted based on measured data from human walking [7]. In the simulation environment the shank is fixed and the foot acts like a pendulum around the lower end of the shank. To ensure a realistic simulation, joint angles are used as desired trajectories, while an external torque is introduced to model the floor reaction forces. The system is modeled by two equations of motion: One representing the drive sided dynamics and one depicting the foot mechanics. The angle ψ between foot and shank is the controlled variable of the system and is compared to the reference angle. The output of the controller represents the motor torque. In the left part of Figure 3 the angle of the ankle-foot prosthesis and of the human data is presented. Here, the trajectory provided by the proposed prosthesis deviates only slightly from human data. The variation at the end of P2 and beginning of P3 implies that the angle between shank and foot is greater than the human data. The required mechanical power, given in the right part of Figure 3 shows the ankle power of the prosthesis and of the human data. The peak appears at the time of push off, in which the prosthesis mimics the catapult effect and thus the propulsion in walking direction. In comparison to the human data the variations occur because the prosthesis mimics the human walk but has a different assembling i.e. the mechanical stops. The simulation is a first and good approach to verify that the design connected with the operational strategy is able to mimic human gait.

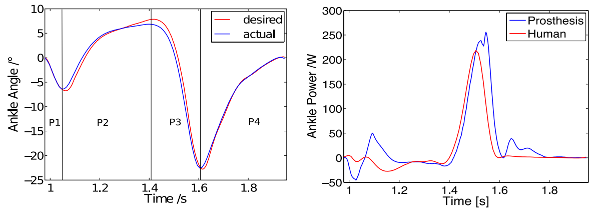

Figure 3: (a) Ankle angle; (b) Ankle power

6 Conclusion

The main innovation of the proposed concept is the interaction of the SEA with a partially active parallel spring, which is used for energy storing and for creating a counter torque to the external torque. In contrast to existing concepts using a parallel spring as the one in [1], this spring is designed to exploit stored energy specifically. The energy is stored during P2 and used to support the actuators during P3. The force applied by motors to the heel during P2 is decreased because the parallel spring creates up to 25% of the required ankle torque. The peak motor torque in simulation exceeds the nominal motor torque for a negligible period of time. This might be provided by overloading the drives. The simulations reveal that it is possible to nearly imitate the biomechanical model with a realistic power requirement using a PID controller. In reality, the control strategy should be extended to a hybrid control system with a position and force controller for realization and better usability.

7 Format

An oral presentation is preferred by the authors.

8 Open Questions

Further simulations should consider the interaction of human and prosthesis by applying more enhanced human model. Regarding the motor torque, the overloading capabilities of the drives have to be considered for implementation.

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