Design and Characterization of a Biologically Inspired Quasi-Passive Prosthetic Ankle-Foot

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***Abstract*—By design, commonly worn energy storage and release (ESR) prosthetic feet cannot provide biologically realistic ankle joint torque and angle profiles during walking. Additionally, their anthropomorphic, cantilever architecture causes their mechanical stiffness to decrease throughout the stance phase of walking, opposing the known trend of the biological ankle. In this study, the design of a quasi-passive pneumatic ankle-foot prosthesis is detailed that is able to replicate the biological ankle's torque and angle profiles during walking. The prosthetic ankle is comprised of a pneumatic piston, bending spring and solenoid valve. The mechanical properties of the pneumatic ankle prosthesis are characterized using a materials testing machine and the properties are compared to those from a common, passive ESR prosthetic foot. The characterization spanned a range of ankle equilibrium pressures and testing locations beneath the foot, analogous to the location of center of pressure within the stance phase of walking. The pneumatic ankle prosthesis was shown to provide biologically appropriate trends and magnitudes of torque, angle and stiffness behavior, when compared to the passive ESR prosthetic foot. Future work will focus on the development of a control system for the quasi-passive device and clinical testing of the pneumatic ankle to demonstrate efficacy.**

I. Introduction

The mobility of transtibial, or below knee, amputees is limited by the design of common, passive ankle-foot prostheses. The muscles that span the biological ankle joint provide the majority of mechanical power during walking. Consequently, the loss of this joint causes these individuals to walk up to 40% slower [1] and expend at least 20% more metabolic energy [2], when compared to non-amputees; a metabolic burden typically associated with carrying 15 kg [3].

Historically, energy storage and release (ESR) prosthetic feet have been developed to reduce some of the aforementioned deficits associated with transtibial amputations. These prosthetic feet typically consist of an anthropomorphic carbon fiber leaf spring, where the leaf spring is cantilevered from the heel anteriorly towards the toe. The purpose of the leaf spring is to store elastic energy as the foot is dorsiflexed during the stance phase of walking, the region of the gait cycle when weight is borne by the leg [4]. Ideally, this energy is returned to the wearer at terminal stance phase, when the ankle plantarflexes to propel the wearer forward.

As a result of the cantilever nature of the leaf spring in the design of ESR prosthetic feet, they do not provide the biologically appropriate torque-angle and stiffness properties during walking. That is, as the center of pressure moves anteriorly during stance phase, the stiffness of the cantilever leaf spring decreases exponentially. This opposes the trend known to occur in the biological ankle joint. The stiffness of the biological ankle joint is known to increase linearly during the dorsiflexion region of stance phase [5]. Therefore, the anthropomorphic cantilever design of ESR feet results in non-biological mechanical behavior, likely contributing to the gait deficits of transtibial amputees.

Researchers have previously designed novel passive and quasi-passive[1] prosthetic feet to address the limitations of current technology. Hansen and Nickel designed an ankle-foot prosthesis to increase balance during walking and standing [6]. To this end, their device incorporated a locking mechanism to transition between two stiffness modes to provide the appropriate kinematic rollover shape observed during walking and standing. Additionally, Collins and Kuo developed a quasi-passive ankle prosthesis that recycles the impact energy from heel contact and returns it during push off [7]. This prosthesis technology was shown to increase ankle push off work and decrease the metabolic energy consumed during walking, when compared to walking with a conventional prostheses. Lastly, recently developed powered ankle prostheses have been shown to normalize transtibial amputee gait characteristics [8]. Previous work in the development of novel ankle-foot prostheses is encouraging and underscores the importance of prosthesis stiffness properties, as well as the significance of energy returned to the wearer during locomotion. Unfortunately the clinical impact of such work has been limited by mechanical complexity; non-biological, cantilever stiffness behavior; as well as substantial mass.

The purpose of this study is to introduce the design of a novel ankle-foot prosthesis that is able to replicate the biological ankle's natural stiffness behavior. The intent of this work is to provide a customizable, lightweight and

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[1] In this work, *quasi-passive* refers to a device that uses a small amount of energy, typically electrical energy to power a microcontroller, but does not provide any net-positive mechanical energy.

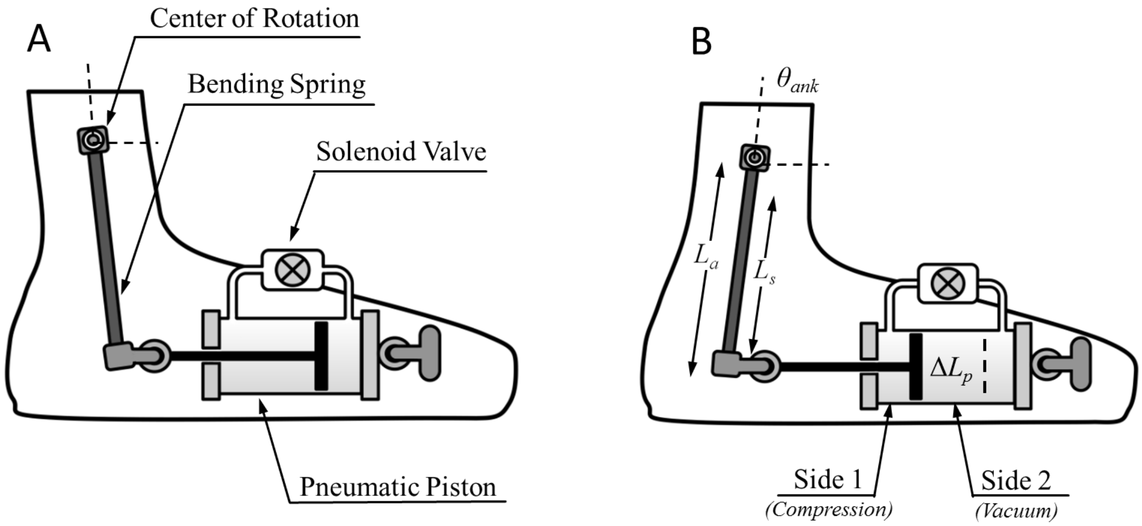

Fig. 1: A) Pneumatic quasi-passive ankle-foot prosthesis mechansim shown with major components depicted. B) Relevant variables are shown. As the ankle is dorsiflexed, the total piston length is elongated and pressure increases in the proximal side of the piston (Side 1). When the valve is opened, the pressure between the two piston sides is equalized, and the ankle rotates freely.

simple design that is able to reduce the locomotory deficits of transtibial amputation. The governing mechanism equations are introduced and the mechanical design of the pneumatic quasi-passive ankle is detailed. The technology is characterized using a materials testing machine and the results are compared to the mechanical properties of a common ESR prosthetic foot.

II. Design

A. Mechanism Architecture

In this section, the design of the pneumatic quasi-passive ankle is detailed. The overarching design goals were to develop a customizable, lightweight ankle prosthesis mechanism that replicated the stiffness characteristics shown recently for the biological ankle joint during walking. To this end, a pneumatic piston with solenoid valve was used in series with a mechanical bending spring (Fig. 1A). This architecture exhibited the biologically appropriate stiffness characteristics with less mass than coil springs that are used traditionally. During operation, the ankle's torque-angle relationship is governed by the force through the piston and lever arm of the bending spring (Fig. 1B). Upon heel contact, the solenoid valve is closed when the foot is fully plantarflexed. As the foot dorsiflexes, energy is stored in the change in gas volumes and flexing of the bending spring. To obtain an expression for the ankle's torque-angle relationship, the gas dynamics were modeled. Assuming negligible transfer of heat between the piston and the surroundings (i.e. adiabatic process), the pressures before and after the adiabatic process are governed by

$$P_A V_A^{\gamma} = P_B V_B^{\gamma} \tag{1}$$

where P and V denote absolute pressure and volume; subscripts A and B denote after and before the adiabatic process, and γ is a constant, known to be 7/5 for diatomic gasses most commonly found in air. The equation for piston force, equation 1 is considered for both the compression and vacuum sides of the piston including area consumed by the piston shaft,

$$F = P_{amb} a_s + \frac{(a - a_s) P_{B1} V_{B1}}{(V_{B1} - a\Delta L_p)^{\gamma}} - \frac{a P_{B2} V_{B2}}{(V_{B2} - a\Delta L_p)^{\gamma}} \tag{2}$$

where P_{amb} is the ambient pressure, a_s is the cross-sectional area of the piston shaft, a is the cross-sectional area of the piston bore, ΔL_p is the change in piston length (stroke), and subscripts *1* and *2* denote the compression and vacuum sides of the piston, respectively. The torque about the ankle, τ, can be determined by

$$\tau = F L_a \cos(\theta_{ank}) \tag{3}$$

where L_s is the length of the total moment arm.

To obtain a corresponding expression for ankle angle, the contribution from the change in piston length must be considered with the contribution from the bending spring flexing

$$\theta_{ank} = \theta_B + \Delta\theta_p + \Delta\theta_s \tag{4}$$

where $\Delta\theta_p$ is angular contribution from the change in piston length, $\Delta\theta_s$ is the angular contribution from the flexing of the bending spring and θ_B is the angle before the process (before ankle dorsiflexion). Using the small angle approximation and the equation governing bending spring displacement the following equations were defined

$$\Delta\theta_p = \frac{\Delta L_p}{L_a} \tag{5}$$

$$\Delta\theta_s = \frac{F L_s^3}{L_a 3EI} \tag{6}$$

where L_s is the length of the bending spring, E is the elastic modulus of the bending spring, and I is the cross-sectional moment of inertia of the bending spring.

The overarching equation that governed the ankle's mechanical properties, torque as a function of angle, was calculated numerically. By substituting equations 5 and 6 into equation 4 and solving for ΔL_p, then substituting into equation 2, the piston force, F, can be used to determine ankle torque with equation 3. Unfortunately, these equations cannot be solved in closed form, and must be solved numerically or through the use of an intermediate variable.

By inspection of equation 2, the stiffness characteristics of the ankle can be tuned by increasing or decreasing the equilibrium pressure within the piston system, P_B. This highlights the customizable nature of the device.

B. Mechanical Design

The pneumatic ankle-foot prosthesis was developed to provide biomechanically appropriate ankle walking kinetics and kinematics in an anthropomorphic envelope. The prosthesis was designed to meet the requirements of a 70 kg person walking at self-selected speeds, while fitting in the biological form factor of a 1st percentile male [9]. These biomechanical requirements were determined from the weight-normalized kinematics and kinetics from a reference dataset [10]. It was assumed that the ankle prosthesis would account for 30% of the volume and mass between the knee and ankle joints, and the entire volume and mass of the foot, accounting for approximately 2.6% of total body mass [10]. The resulting specifications are shown in Table 1. These requirements provided the foundation for the mechanical design of the pneumatic ankle prosthesis.

Fig. 2: A rendering of the pneumatic ankle-foot prosthesis.

Table 1: Prosthesis Design Specifications

Parameter	Value
Range of motion	0.4 radians
Max torque	80 Nm
Mass	< 1100 g
Height	< 178 mm
Foot length	< 232 mm

A pneumatic cylinder and valve in series with a fiberglass bending spring passively stored energy and modulated equilibrium position, where equilibrium position refers to the angular position that causes zero torque The pneumatic cylinder has a bore diameter of 27 mm, a stroke length of 38 mm, and a maximum rated pressure of 1.4 MPa, resulting in a maximum output force of 800 N (model: PSD3, Fabco-Air, Inc., Gainesville, FL). Air flow between the two chambers of the cylinder was controlled with a miniature electric solenoid valve (model: A2015-C203, Gems Sensors & Controls, Plainville, CT). The bending spring was 65 mm long (L_s), 27 mm wide, 6 mm thick in the bending direction, and made from a unidirectional fiberglass composite (model: GC-67-UB, Gordon Composites, Montrose, CO). The bending spring is integrated into a 100 mm (L_a) moment arm that links the pneumatic piston to the ankle joint. The moment arm length selection is a tradeoff between the cylinder bore, length and force with the vertical height of the ankle's center of rotation; all were chosen to fit the aforementioned design criteria. The vertical height of the center of rotation is 130 mm. The cylinder stroke length and moment arm resulted in a 0.36 radian unloaded range of motion, with an additional 0.1 radians of dorsiflexion achieved through loading of the bending spring. The moment arm length and maximum cylinder force resulted in a maximum ankle torque of 80 Nm, which is approximately 400% that typically borne by ESR prosthetic feet [11] and 70% of the maximum torque exerted by the biological ankle of a 70 kg individual [10].

The architecture of the pneumatic ankle prosthesis was designed to handle the loads and torques of level ground walking, while minimizing mass and length in a biological form factor. An aluminum proximal pyramid mount constrained one end of the bending spring and rotated on two angular bearings, each with a load rating of over 1300 N (model: Kit8330, NationSkander California Corporation, Anaheim, CA). The foot is comprised a lightweight plastic sole and a rigid aluminum housing that prevented structural bending of the foot. The mechanical components of the ankle, depicted in Fig. 2, have a total mass of 1040 g, a height of 177 mm, and a foot length of 188 mm, all within the design specifications for a 1st percentile male.

III. Methods

To obtain an understanding of how the mechanical properties of the pneumatic ankle prosthesis compare to common commercially available ESR prosthetic feet, both were characterized in a materials testing machine.

A. Testing of the Pneumatic Prosthesis

The pneumatic quasi-passive prosthesis was secured to an aluminum testing rig that was fastened to the high-capacity frame of the testing machine (model: 1125 with A30-33 load cell, Instron, Norwood, MA). The testing machine was chosen for its ability to apply precise displacements while synchronously measuring resultant forces. The starting angle of the pneumatic ankle was set to maximum plantarflexion (0.2 radians, mechanically adjustable) to characterize the full dorsiflexion range of motion. During testing, a servo-ram was lowered, applying a displacement to the pneumatic foot through a high force roller (32 mm radius) mounted to the end of the ram. The roller was used to eliminate sliding friction at the contact between the servo-ram and the foot as the ram traveled linearly. The perpendicular distance of the ram to the center of rotation of the ankle was varied, analogous to differing centers of pressure (COP) distances beneath the foot. The ankle was characterized at ram distances of 102 mm, 122 mm, 145 mm, 168 mm and 185 mm from the center of rotation anteriorly. These distances were chosen because they span the testable range of the foot, accounting for translation (rolling) of the ram along the anterior-posterior axis of the foot as the ram lowered and the ankle dorsiflexed. Two equilibrium pressures were tested at each ram distance, 310 kPa and 413 kPa. At the ram distance of 145 mm, five equilibrium piston pressures were tested, 0 kPa, 103 kPa, 206 kPa, 310 kPa and 413 kPa (gauge pressure). A single trial was acquired at each ram distance and equilibrium pressure, with force data acquired at 10 Hz. During testing, the ram lowered with a velocity of 20 mm/min and the test was terminated when the torque about the ankle reached approximately 70 - 80 Nm.

B. Testing of the Commercial ESR Prosthesis

An identical protocol was used to test the ESR prosthetic foot (model: Seattle Low Profile, 27 cm, category 3; Trulife, Poulsbo, WA). The footshell was removed from the foot prior to testing. This was to eliminate substantial local compression of the foam footshell not likely to be experienced during locomotion. The foot was secured to the testing rig at a neutral angle (0 radians) and the servo-ram applied a displacement to the foot and recorded the force. Five ram distances were tested spanning the length of the foot; the ankle was characterized at ram distances of 53 mm, 73 mm 109 mm 129 mm and 154 mm from the ankle's center of rotation. Since the ESR has a fixed ankle (no mechanical revolute joint), an axis was determined to be the ankle's center of rotation and was consistently used in the analysis.

C. Data Analysis

The force and displacement measurements were converted to ankle torque, τ, and angle, θ_{ank}, using the geometry of the measurement setup and prosthetic feet (identical geometric analysis described in [12]). All data were low-pass filtered using a bi-directional third order Butterworth filter with a cutoff frequency of 5 Hz. For the pneumatic prosthetic foot, MATLAB (The Mathworks, Natick, MA) was used to calculate the stiffness, $d\tau/d\theta_{ank}$. A second order polynomial was fit to four points surrounding each point, and the polynomial coefficient was used to quantify the derivative [13]. For the ESR prosthetic foot, linear regression was used to estimate the stiffness because the stiffness did not vary with angle. All data are reported as mean ± standard deviation.

IV. RESULTS

A. Mechanical Behavior and Equilibrium Pressure

The pneumatic ankle prosthesis was tested at five equilibrium pressures, ranging from 0 kPa to 413 kPa at a ram distance of 145 mm. As the pressure was increased, the torque increased as a function of angle (Fig. 3). The torque varied from 0 Nm to approximately 30 Nm – 80 Nm, depending on the equilibrium pressure tested. The equations describing the pneumatic ankle were used to predict the torque as a function of angle. The model predicted data are shown for comparison for the same equilibrium pressures. The trends in the data generally agree, with greater deviation at low equilibrium pressures and dorsiflexed ankle angles. At 0.1 radians dorsiflexion, the mean difference in torque was 5.8 ± 1.2 Nm, when compared across equilibrium pressures.

B. Mechanical Behavior and Ram Distance

The torque and angle behavior were quantified at several ram distances for both the pneumatic ankle prosthesis and the ESR ankle prosthesis. The torque-angle relationship of the pneumatic ankle prosthesis did not vary with ram distance, shown at an equilibrium pressure of 413 kPa (Fig. 4A). The torque increased to approximately 45 Nm at 0.15 radians dorsiflexion with a total of approximately 0.3 - 0.4 radians displacement. Thus, the stiffness of the pneumatic ankle prosthesis did not vary with ram distance. The torque-angle relationship for the able-bodied ankle is shown for reference.

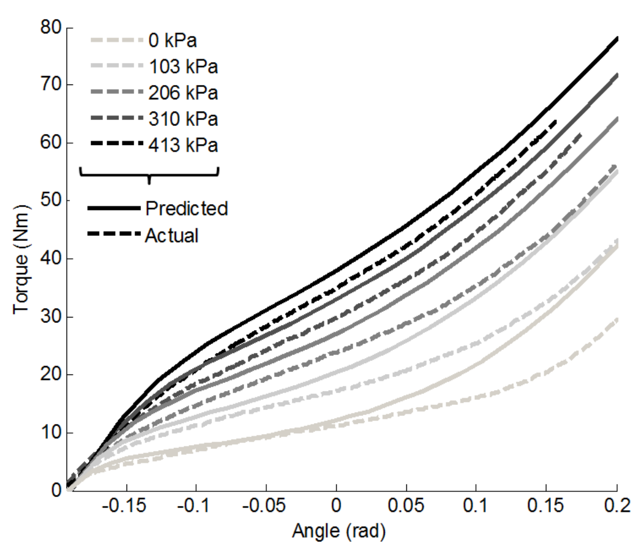

Fig. 3: Actual and model predicted values of torque as a function of angle. The color denotes equilibrium gauge pressure and the dash/solid represents actual or predicted data. Note the model agreement increases with displacement. This is likely a result of the simplified bending spring modeling and adiabatic assumption.

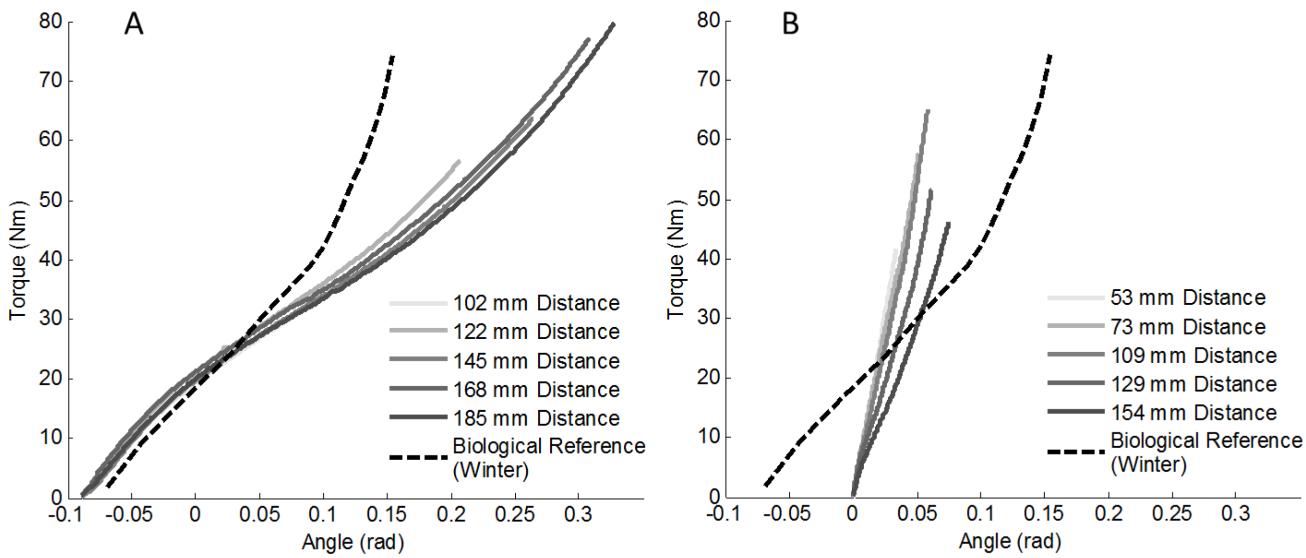

Fig. 4: A) Torque as a function of angle for the pneumatic ankle prosthesis (413 kPa), shown with biological reference data. Note the five distances tested agree in the torque-angle space, a result of the articulated mechanical design. This relationship can be adjusted by varying the equilibirum pressure in the cylinder. B) Torque as a function of angle for the ESR prosthetic foot, shown for the biological reference. Note that the stiffness of the ESR foot decreases with increasing distance tested.

The torque-angle relationship of the ESR prosthetic foot varied with ram distance (Fig. 4B). As a result, the angular displacement required to generate 40 Nm varied from 0.03 – 0.06 radians (mean 0.045 ± 0.015 radians), depending on ram distance. As ram distance increased from 53 mm to 154 mm (i.e. the ram was translated anteriorly), the stiffness of the ESR prosthetic foot decreased by 51%. Thus, the stiffness of the ESR foot is a function of the ram distance, analogous to COP location during walking. The stiffness of the ESR prosthetic foot decreases during the stance phase of walking. When compared to the reference dataset for the biological ankle, the stiffness of the ESR is substantially greater severely limiting the magnitude of angular deflection of the ESR prosthetic foot.

C. *Ankle Angle, COP Location and Stiffness*

The measured stiffness of the pneumatic ankle prosthesis was compared to the stiffness of the ESR prosthetic foot, and recently published biological ankle stiffness values [5]. To fully understand how the pneumatic ankle prosthesis is different than the conventional ESR prosthetic foot, the stiffness data were compared as a function of ankle angle and normalized COP location (Fig. 5). Normalized COP location was defined as the ram distance divided by the total foot length. The stiffness of the pneumatic ankle prosthesis (413 kPa) increased by a factor of three, following similar trends to those observed in biological data. Work estimating ankle impedance during walking showed ankle stiffness increasing by a factor of four as the ankle was dorsiflexed and the COP translated anteriorly (COP data obtained from an open dataset [14]). It should be noted that the negative linear trends of pneumatic ankle COP and ankle angle is an artifact of the testing apparatus and protocol.

The behavior of the ESR prosthesis opposed the magnitude and trends of biological stiffness data. The stiffness of the ESR prosthetic foot was at most 11 times greater than the biological ankle, with the stiffness decreasing as ankle angle and center of pressure increased, opposing biological trends.

V. Discussion

This study introduced a novel pneumatic ankle-foot prosthesis design that is capable of providing biologically appropriate kinetics and kinematics in a simple, lightweight design. Additionally, the pneumatic ankle prosthesis was designed to provide biomimetic torque-angle behavior, independent of center of pressure location. This work was motivated by the non-biological behavior of common, commercial ESR prosthetic feet. Such prostheses use an anthropomorphic cantilever beam design that inherently causes the stiffness to decrease as the center of pressure travels anteriorly during the stance phase of walking. The pneumatic ankle prosthesis and a commercial ESR prosthetic foot were characterized and the mechanical behaviors were compared to biological reference data.

The error in the model predicted torque-angle relationship of the pneumatic ankle prosthesis decreased with increasing equilibrium pressure (Fig. 3). This discrepancy may be the result of non-ideal behavior from the bending spring within the mechanism or the idealized adiabatic assumption. The mathematical model of the pneumatic mechanism was used to design the physical specifications, and ideal behavior is not required for use. The error could likely be reduced if the bending spring were characterized, rather than assuming ideal behavior. Furthermore, any discrepancy in tubing volume between the model and the physical system may cause error.

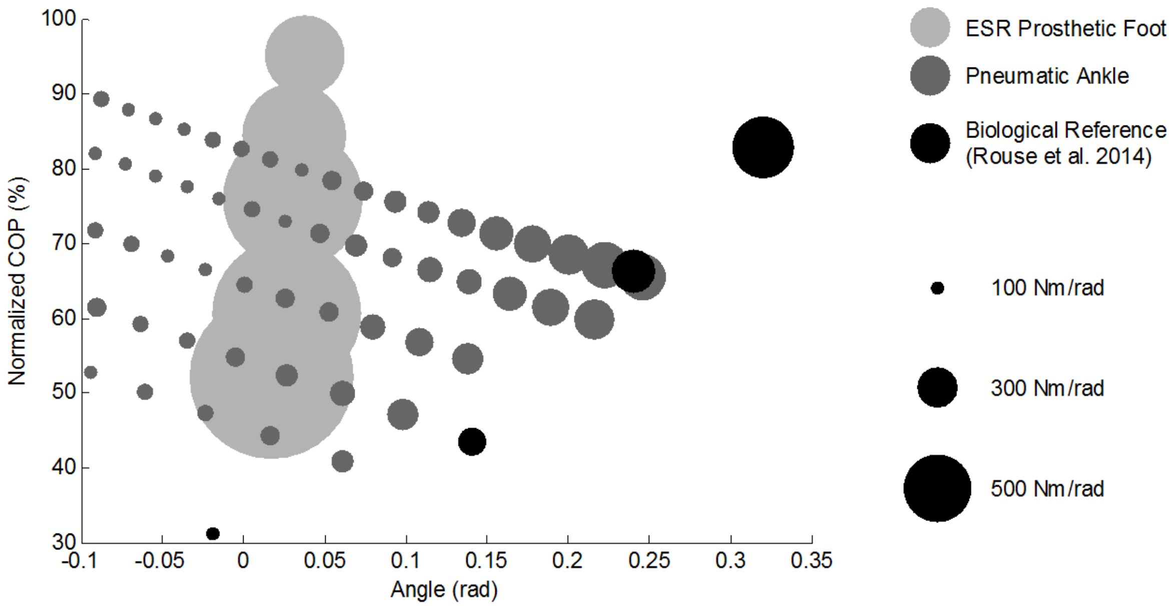

Fig. 5: Stiffness shown as a function of ankle angle and normalized center of pressure (COP). The size of the markers denotes the magnitude of the ankle stiffness at the respective ankle angles and COP location during walking. COP is normalized by foot length, to negate the effect of the different designed lengths of the prosthetic feet. Note that because the stiffness of the pneumatic ankle is only a function of angle, movement along the vertical axis does not affect stiffness magnitude. Corrspondingly, for the ESR foot, stiffness is only a function of COP location, and movment along the horizontal axis does not affect stiffness magnitude. Human data reported from Rouse et al. [5], shown for a 70 kg individual.

The dependence of the pneumatic ankle's mechanical behavior on piston equilibrium pressure presents the opportunity to conveniently customize the pneumatic prosthesis for each individual. The trends of the torque-angle relationship may be modified (Fig. 4A) by increasing or decreasing the equilibrium pressure, see (2). Commercial ESR prosthetic feet are not able to have their mechanical behavior modified once constructed. Such a convenient and reversible mechanism for mechanical behavior modification presents the possibility for a single prosthesis to provide biologically appropriate mechanical behavior across a range of walking speeds, activity levels and load carriage.

The pneumatic ankle prosthesis and the ESR prosthesis had opposing stiffness trends. The pneumatic ankle prosthesis was designed for torque to vary only as function of ankle angle, with stiffness increasing as the ankle is dorsiflexed (Fig. 4A). This design was based on recently published biological data, shown for reference in Fig. 5. As a result of the cantilever nature of ESR prosthetic feet, their stiffness properties decrease as COP moves anteriorly during the stance phase of walking. To compare stiffness at varying locations of COP and ankle angle, the pneumatic ankle stiffness is approximately constant across the vertical axis (i.e. COP location). Conversely, the stiffness of the ESR prosthesis is approximately constant across the horizontal axis (i.e. ankle angle).

The biologically inspired behavior of the pneumatic ankle prosthesis may be important for earlier release of stored mechanical energy for push off. The increasing stiffness of the pneumatic ankle prosthesis provides the stored energy earlier for the wearer to push off during walking. When stiffness increases as a function of angle, more energy is stored per unit of angular displacement as the ankle is dorsiflexed. Therefore, as the ankle begins plantarflexing during push off, the energy is returned back to the user, with most of the energy stored at the greater angles of dorsiflexion. The motivation for this design is that the earlier return of stored energy may reduce the metabolic energy consumed by the wearer during walking.

The use of the solenoid valve provides the ability to arbitrarily set the equilibrium position of the pneumatic ankle prosthesis. In other words, the solenoid valve may essentially provide the function of a clutch, commonly incorporated in biologically inspired mechanisms [15, 16]. When the solenoid valve is opened, air is free to pass between the piston sides, allowing the joint to rotate freely. Thus, the position at which the solenoid valve is closed becomes the equilibrium position until the valve is opened again. This permits dorsiflexion of the ankle during the swing phase of walking, providing required toe clearance. Although the development of a control system for the pneumatic ankle prosthesis was outside the scope of this work, the use of the solenoid valve has advantages for the wearer.

This study set forth the design and characterization of a pneumatic ankle-foot prosthesis. The design was shown to have customizable, biologically inspired mechanical properties, contrary to commercial ESR prosthetic feet that have opposing trends and magnitudes when compared to biological reference data. Future work will focus on the development of a control system for the pneumatic prosthesis and clinical testing of gait characteristics and metabolic expenditure during walking.

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