Redefining Prosthetic Ankle Mechanics

Non-Anthropomorphic Ankle Design

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***Abstract*— The moment transferred at the residual limb socket interface of transtibial amputees can be a limiting factor of the comfort and activity level of lower limb amputees. The high pressures seen can be a significant source of pain, as well as result in deep tissue damage. The compensation of the sound limbs causes an asymmetrical gait which can be a contributor of early onset osteoarthritis in the sound limbs. It has been shown that the moment transferred with conventional passive prostheses can be lowered in magnitude by aligning the tibia with ground reaction forces, but this limits the effectiveness of the device. With recent powered prosthetics designed to mimic the missing limb, power can be injected into the gait cycle, but can also be limited by this pressure threshold. This paper shows the results of calculations that suggest that altering the prosthetic ankle mechanism can reduce the socket interface moments by as much as 50%. This supports the development of an active non-anthropomorphic ankle prosthesis which reduces socket interface moments while still injecting substantial power levels into the gait cycle.**

***Keywords*— powered ankle prosthesis; transtibial; lower limb amputee gait; non-anthropomorphic design**

I. INTRODUCTION

With the prevalence of dysvascular disease and the general aging of the US population, it is estimated that the percentage of people living with major lower limb amputations is likely to double from 1 in 500 as of 2005 by 2050 [1][2]. In a 2001 survey it was reported that 51% of amputees experience pain while walking [3][4]. This pain and discomfort can lead to rejection of the prosthesis. This is an indicator that current approach to prosthesis design is not fulfilling the needs of amputees.

Significant technological advancements over the past decade have made the realization of a new class of intelligent prostheses for lower limb amputees possible, compensating for the lost function and power of a missing limb [5–7]. The goal in the design of these lower limb prostheses has been to mimic the lost limbs as exactly as possible, assuming that the socket-limb interface to the amputee is rigid and comfortable. Since the socket is supported by soft tissue, the connection to the amputee is far from ideal and results in gait abnormalities for the amputee as well [8]. During normal walking, the forces and moments generated in the prosthesis must be transmitted through the socket to the soft tissue of the residual limb. As the tissue compresses, the load is transferred to the residual limb causing uneven pressure distributions that are very high, are a source of pain [9], and can cause further damage to the residual limb [10]. This pain and discomfort may limit the functionality of the prosthesis in terms of walking speed, stride length, and maximum push-off forces, requiring the amputee to compensate with their intact limb. Amputees are twice as likely to have pain in their intact knee [8], and seventeen times more likely to have osteoarthritis than age matched non-amputees [11].

Many health issues faced by amputees can be attributed to the fact that the current lower-limb prostheses create abnormal loading conditions on the residual limb. The ground reaction force and resulting moments must be transmitted through the socket-limb interface rather than through bone transfer as in an intact limb. Ways to reduce the magnitude of the load with an active prosthesis have not been explored. *Altering the alignment of the ground reaction forces with the residual limb throughout stance can reduce the magnitude of the moment loading on the residual limb.*

The design principle for lower limb prostheses has always been to replace the form and function of the lost limb as

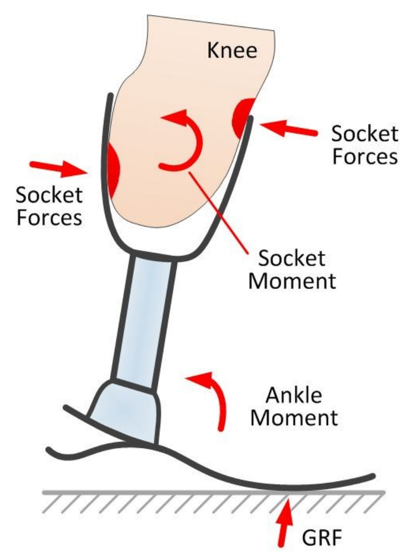

Fig. 1. Transtibial socket loading during gait with highlighted regions of high pressure.

closely as possible [12]. However, such an approach does not account for the altered anatomy of the amputee and the connection of the prosthesis to the individual. Large moments caused by loading of prosthetic feet must be transmitted through the socket interface. The most common sockets used are the patellar tendon bearing (PTB) sockets [13]. As a result, abnormal loading conditions on the residual limb compress the tissue against the bone. A simplified perspective is to view the socket interface in terms of three-point bending. Although the contact surface of the socket over the residual limb is continuous, the moment loading creates high pressure areas as seen in three-point bending and illustrated in Fig. 1. This can be painful and cause further damage to the residual limb [14].

To reduce the pressures on the residual limb at the socket interface, the effect of increasing the compliance of the prosthetic foot reducing socket loading has been studied [12]. However, this approach also reduces the range of motion in the affected knee during stance which lowers the maximum elevation of the person's center of mass during mid-stance [12][15]. Another method used to even out the pressure distribution is adding compliance at the socket interface using thicker, low-compliance gel liners [16]. The added compliance reduces the peak pressures, but it limits the energy return from the prosthesis causing the sound side to compensate even more. Further, they increase thermal issues and decrease the user's sense of stability and sensory feedback which in some cases can result in higher ground reaction forces [17]. Other studies have been done observing the effects on load transfer due to changes in pylon stiffness [18].

Another common method in practice to reduce maximal residual limb pressures in late stance is through static alignment of the prosthesis foot in relation to the residual limb, shifting the prosthetic connection point anterior in the sagittal plane with special hardware, which effectively reduces the moment arm [19]. This shift is constant throughout the gait cycle however, which decreases the ability to push off during rollover. It has been shown that while this decreases the maximum moment during push-off, it actually increases the negative moment following heel strike [20]. The approach reduces the socket moment and makes use of the prosthesis more comfortable. However, the sound side is still forced to compensate for the decreased performance of the disabled limb.

In this paper, we develop the concept of a non-anthropomorphic active lower limb prosthesis. The approach actively realigns the prosthesis to redirect ground reaction forces to be more in line with the tibia which lowers the moment needed to be transferred, with a slightly altered gait. As a result, it may be possible to inject healthy power levels into the gait cycle in order to restore rest of body biomechanics in a more comfortable manner that is more appropriate for an amputee's altered anatomy. Our preliminary studies (detailed in *Section II*) have shown that altering the knee angular trajectory during the gait cycle can realign the tibia with the ground reaction forces, lowering the peak moments during gait.

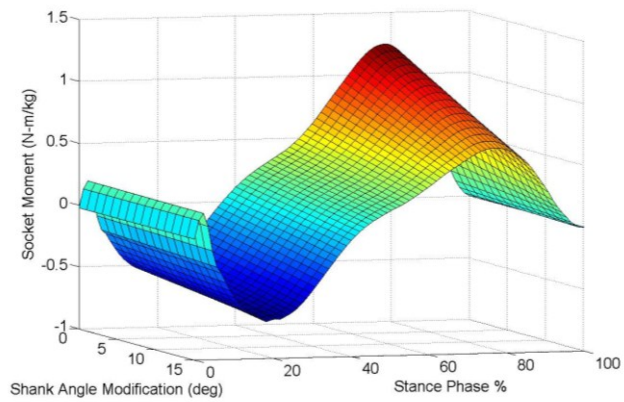

Fig. 2. Socket interface moment reduction by aligning the shank with the CoP. Peak moments were reduced from 1.25 to 0.64 Nm/kg (a 50% reduction) at 60% of the gait stance phase.

II. Altered Biomechanics Study

Data from one able-bodied, non-amputee subject (29 years of age, 70.2 kg, 1.67 m) was collected and analyzed to examine the possible effects of altering the tibia angular trajectory on mid-tibia moments. Able-bodied data is used instead of amputee data in order to maintain able-bodied biomechanics throughout gait in the intact limbs, including the foot to ground angular relations in order to keep the center of pressure and ground reaction force trajectories unaltered. Tibia angular trajectory was then altered with the assumption that the data came from an amputee, and a mechanism between the tibia and foot existed allowing this motion. The resulting mid-tibia moment was then calculated. It is noted that with this method, knee and hip moments in the altered limb would be altered as well, however only mid-tibia moments are examined since they are a source of high pressures in the socket interface of an amputee.

The experiment took place in the Biomechanics Laboratory at the University of Massachusetts, Amherst. The

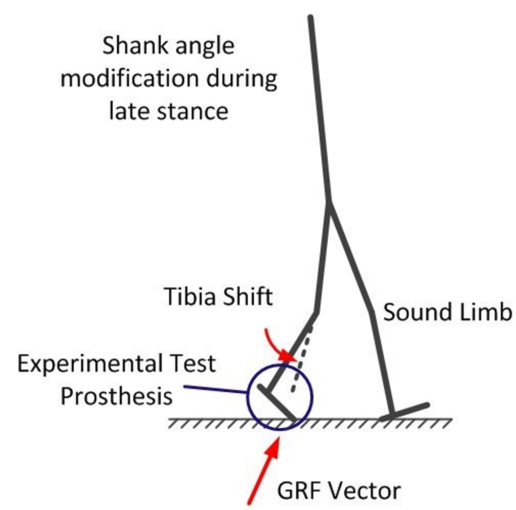

Fig. 3. Illustration of concept showing shank angle modified during late stance to reduce moment transfer at the socket residual limb interface.

data collection used an eight-camera Qualisys Oqus 3-Series optical motion capture system operated by Qualisys Track Manager software (Qualisys, Inc., Gothenberg, Sweden) to sample test subjects fitted with reflective infrared tracking sensors, at 240Hz. Ground reaction forces were recorded with a floor mounted strain gauge force platform (OR6-5, AMTI, Inc. Watertown, MA, USA). The force platform recorded the shear forces in two directions and normal forces. The calibration markers were placed at the following anatomical features to reconstruct the bone structure during data processing: 1st and 5th metatarsals, medial and lateral knee joint as well as ankle malleoli, and the greater trochanters in order to reconstruct points of rotation. Four tracking markers were fixed to each foot, shank, and thigh, as well as the hip segment throughout all testing to track the trajectories of each segment. The data were processed using Visual 3D v4 software (C-Motion, Inc, Rockville, MD, USA) to calculate all joint positions, velocities, moments, and power. MATLAB (MathWorks, Natick, MA) was used to analyze and perform the tibia altering calculations.

In Fig. 2, it can be seen that the calculated moment transferred through the socket-limb interface could be decreased by almost 50%. This decrease in moment assumes that the whole body biomechanics from the affected knee up remain unchanged, as well as the foot rotation in relation to the ground which corresponds to the center of pressure trajectory used to make the calculations. The calculations were made by examining the magnitude and direction of the ground reaction force vectors of collected able-body data in relation to mid-tibia with the altered trajectory. This represents moments that would be seen at the socket interface of an amputation halfway between the malleoli and knee if power levels seen in healthy gait are transmitted, corresponding to an ideal active prosthesis.

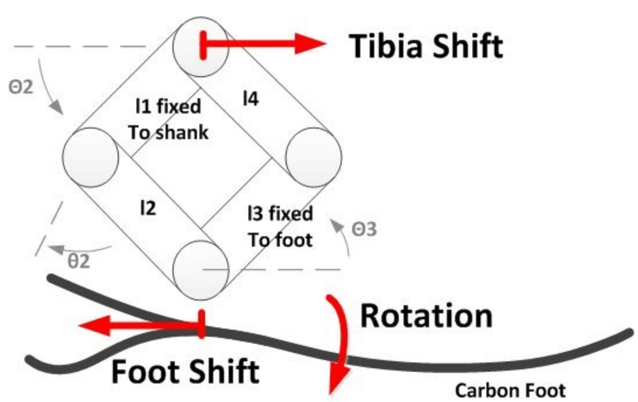

Fig. 4. Four bar mechanism shown with link lengths l_1, l_2, l_3, l_4 fixed joint angles θ_1, θ_3, used to attain desired rotation and translation of the foot, and dynamic angle θ_2 which dictates the motion of the device.

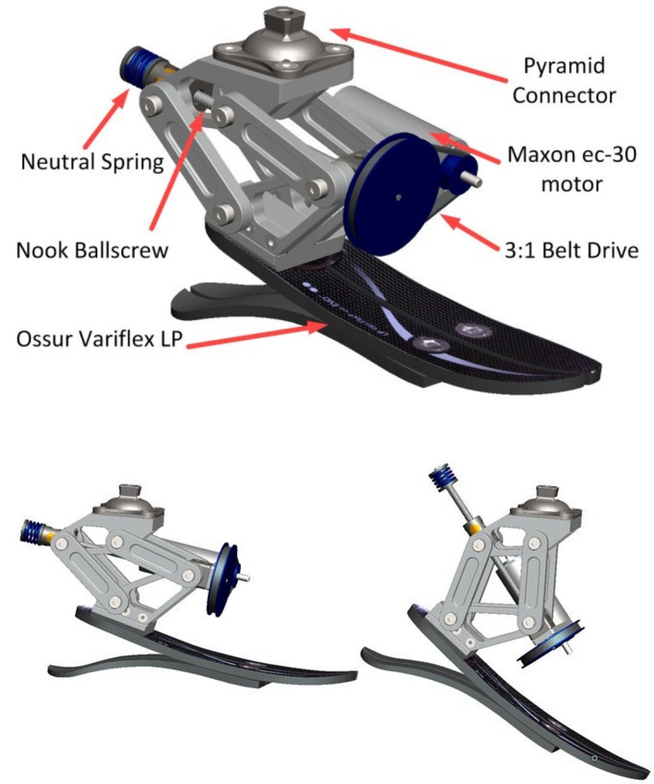

Fig. 5. (top) Test apparatus mechanical assembly showing main mechanical components. (bottom) Test apparatus shown at the neutral standing position (left), and at the shifted axis of rotation position (right).

III. NON-ANTHROPROMORPIHC PROSTHESIS DESIGN

A. Mechanical Design

The purpose of this design is to create a powered prototype prosthesis that alters the tibia angle throughout the stance phase of walking, while maintaining rest of body biomechanics as best as possible (as done in the Fig. 2 moment calculations and illustrated in Fig. 3). When the tibia angle is altered throughout stance, the center of rotation of the foot translates since the prosthetic foot trajectory is also designed to maintain natural biomechanics rotating in relation to the ground as a healthy foot would during normal gait. This ensures proper trajectory of the center of pressure. The overall height of 16 cm is comparable to active and semi-active prostheses currently on the market, which will accommodate a majority of test subjects. Lastly, the mechanism is designed to allow different amounts of total translation of the center of rotation between tests, which effects the maximum moment reduction at the socket interface. The tibia shift is expressed in terms of the translated distance of the prosthetic ankle's center of rotation in the design section since between test-subjects, different tibia lengths will result in different amounts of angular shift. When analyzing results, translational shift will be expressed in terms of degrees of tibial shift in order to normalize the data between subjects in future testing.

An optimized actuated four-bar linkage (Fig. 4) was designed to implement the desired functions. Links 1 and 2 are fixed to the shank and foot respectively. The amount of translation of the foot axis of rotation depends on the length l_2 and amount this link rotates during actuation. The amount of rotation of the foot depends on the l_1 to l_3 ratio when links 2 and 4 are equal. At the neutral position during standing, the center of rotation needs to line up on center with the tibia, so the foot is in a natural position when not walking, providing standing stability. Since one of the design necessities is to maintain foot rotation as it would occur naturally, the flexion during standing and stance of a conventional carbon prosthetic foot is accounted for in the design calculations in order to ensure proper heel lift and push-off.

With these design constraints, the link lengths and fixation angles relative to the foot and socket were solved for different amounts of rotational axis translation minimizing the sum of the link lengths, where link lengths 1 and 3 attached to the socket and foot and are varied, and link lengths 2 and 4 are constant in all three designs. This minimization was done in order to decrease the overall size and weight of the mechanism. Amounts of 3, 6, and 9 cm of translation were chosen for initial testing since they fall into the 5 to 15 degree tibia shift range for subjects less than 1.85 m tall. The same prosthesis will be able to be tested on taller subjects; however less tibia angular shift will be seen. Table 1 contains all link lengths and angles of attachment that were solved for.

TABLE I. DESIGN PARAMETER VALUES FOUND TO OBTAIN DESIRED AXIS TRANSLATION FOR A 1.67 M TALL SUBJECT.

Link lengths (cm)				Fixed Angles (deg)		Axis Translation (cm)
l1	l2	l3	l4	θ1	θ3	
4.50	6.36	6.55	6.36	0.00	22.60	3
4.53	6.36	6.36	6.36	6.81	23.32	6
4.85	6.36	6.36	6.36	21.80	29.69	9

The device as designed for 9 cm of shift can be seen in the top of Fig. 5 with the major mechanical components labeled. The movement of the linkage translating the axis of rotation during actuation is illustrated in the bottom of the figure. It can be seen that as the foot rotates, it shifts posteriorly bringing the tibia more in line with the center of pressure.

The links are fabricated components designed to the strength of 7050 aluminum. The links are separated at the pivot points by low friction washers, and have 6mm ID needle bearings that pivot around shoulder bolts. The ball screw nut carrier and thrust bearing carrier are machined from cold-rolled steel, which will act as the shoulder bolt attachment points for the link 1-2 pivot, and link 3-4 pivot respectively. The ball screw thrust bearing is a single row ball bearing since the actuator will always be under tension. Connection to the amputees socket will be done via a custom machined pyramid connector, which enables height extension with standard adjustable pylons to accommodate different subjects.

B. Actuation

The foot-ankle prosthesis is actuated with a ball screw mounted on pivot points centered with the rotational points between links 1 to 2, and links 3 to 4. To avoid imposing a moment on the ball screw, symmetric linkages are on either side of the ball screw in the sagittal plane. This device is only intended for studying its effects during stance of normal walking, and has no need to lift up on the toe. It therefore will only put tensile stresses on the ball screw due to the nature of the linkage. The ball screw diameter was minimized since buckling is not an issue, which is 6mm in diameter and 1 mm in pitch manufactured by Nook Industries. The motor is a 200W brushless DC motor (Maxon Motors, EC-30 4-pole) mounted to the side and coupled to the ball screw with a belt drive having a ratio of 3:1, allowing the maximum needed actuation speed of 7 cm/s. A neutral equilibrium spring is mounted at the end of the ball screw which supports the load during standing, as well as during heel-strike so the motor does not need to be stalled, which would cause overheating. The spring has an adjustable equilibrium position to correct for the resulting compression from different subject weights during heelstrike and standing.

C. Sensing

There are three sensors onboard the device in order for control to be implemented throughout the gait cycle. A quadrature encoder mounted on the rear of the motor provides the computation with position and velocity feedback. An inertial measurement unit and gyro monitoring the axis intersecting the transverse and frontal plane provides the computation with tibia orientation to determine phase of gait. Strain gauges are mounted underneath the pyramid connector [5], which indicates ground contact at heelstrike, and also helps to identify phase of gait. These sensors will enable state control to be implemented throughout the gait cycle.

IV. CONCLUSION

The mechanical design presented in this paper is meant to be a test apparatus to be used in the identification of alternative residuum loading techniques, and the effects they have on full body biomechanics, by actively realigning the tibia in relation to the ground reaction force vector during late stance. It can also be used to identify and study the effects of different amounts of gait alterations in order to aid future prosthesis design. Though our calculations suggest that altering the biomechanics of lower limb amputees can have the potential to improve the comfort, and even whole body biomechanics, it is completely unknown how much of an alteration would be possible while maintaining walking stability, as well as how much of an alteration would be accepted by the amputee without feeling like it was too much of a change from what is considered normal. It is highly possible that with these methods, there may be a threshold limiting the amount of alteration to the mechanics when the amputee loses a sense of stability.

Immediate future work involves modeling and simulation of healthy full body biomechanics during a complete gait cycle. The model will consist of a seven link free kinematic chain, with links representing the feet, shanks, thighs, and head arms and torso, all having anthropomorphic length, mass, and inertial properties of a specific subject which biomechanics data can be collected from for model validation.

The model will then be altered to simulate unilateral amputee gait with a passive carbon-spring ankle prosthesis and again with the prosthesis described in this paper, in order to study the effects of the two approaches to prosthesis design on both the moments transferred at the socket residual limb interface as well as whole body biomechanics. These simulations will then be compared to actual biomechanics data of able bodied subjects, and amputee subjects fitted with both conventional passive ankle prostheses, and the prosthesis design described in this paper.

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