Simulation of a Powered Ankle Prosthesis with Dynamic Joint Alignment

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***Abstract*—This paper presents simulations of a new type of powered ankle prosthesis designed to dynamically align the tibia with the ground reaction force (GRF) vector during peak loading. The functional goal is to reduce the moment transferred through the socket to the soft tissue of the residual limb. The forward dynamics simulation results show a reduction in socket moment and the impact on the pelvis and affected-side knee. This work supports further research on transtibial prosthetic designs that are not limited to mimicking physiologically normal joint motions to optimize lower limb amputee gait.**

I. Introduction

A transtibial amputee is connected to their prosthesis through a rigid socket that encapsulates the soft tissue of their residual limb. Moments generated by the prosthesis will be transmitted through the soft tissue of the socket interface, rather than directly to the skeletal structures as in intact-limbs. The highest loading on the residual limb occurs during the stance portion of gait due to the large bending moment, as seen in Fig. 1. The nature of moment transfer at the socket interface causes an uneven pressure distribution and high pressure concentrations on the anterior and posterior of the limb, which can be painful to the amputee and can cause further damage to the local tissue [1].

Patellar tendon-bearing (PTB) sockets are a common type of socket used for transtibial amputees and are designed to load the more pressure tolerant areas of the residual limb [2], [3]; however, high pressure areas remain. This can result in pressure ulcers and deep tissue injury, as well as overloading the sound limb as the amputee develops an asymmetrical gait pattern [4]–[8]. The overloading of the sound limb may explain why a lower-limb amputee is 17 times more likely to have osteoarthritis [9], twice as likely to have pain in their intact knee [10], and report experiencing back pain after amputation [11].

Different methods are used clinically to alleviate the high force concentrations in the socket. Often compliance is added with gel liners to reduce pressures and shear stresses [12]; however, this reduces the amputee's sense of stability and sensory feedback resulting in higher GRFs [13]. The socket is often statically misaligned to the anterior of the foot in an effort to reduce the peak moment transferred during late stance [14], which correlates with a reduction in peak pressures seen in the socket [15]. However, this increases the negative moment seen after heel strike and limits the energy storage and return during push-off [14].

Research on active lower limb prostheses has shown that the unaffected limb of unilateral amputees compensates less when using a powered prosthesis [16]–[18]. However, active prostheses designed to anthropometric constraints are susceptible to the same issues as passive prostheses. The ground reaction forces in the unaffected limb do not change significantly however, showing a slight power burst during very late stance of the affected limb after the second peak, just before push-off [19]. This suggests that the amputee is still compensating with the contralateral limb and is still adjusting their gait to a comfort threshold since the affected side has the same peak ground reaction force during push-off.

Prostheses are usually designed to be anthropomorphic in an effort to replace the form and function of the missing limb [20]. This is done with the assumption that the socket connecting the prosthesis to the residual limb is an ideal, rigid joint, as was modeled by Neptune et al [21]. However, this idealization is not the case since the weight of the individual must be transferred through compliant tissue of the residual limb within the socket. The result is added degrees of freedom within the socket which affect gait dynamics and efficiency.

In this paper, a novel active ankle prosthesis is evaluated in simulations which incorporates Dynamic Joint Alignment (DJA), while injecting power into the gait cycle. This design,

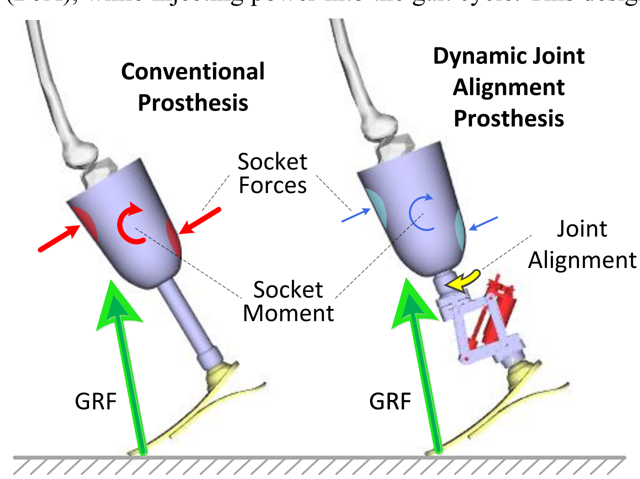

Figure 1. Comparison alignment of the GRF vector with the socket for conventional and Dynamic Joint Alignment prostheses.

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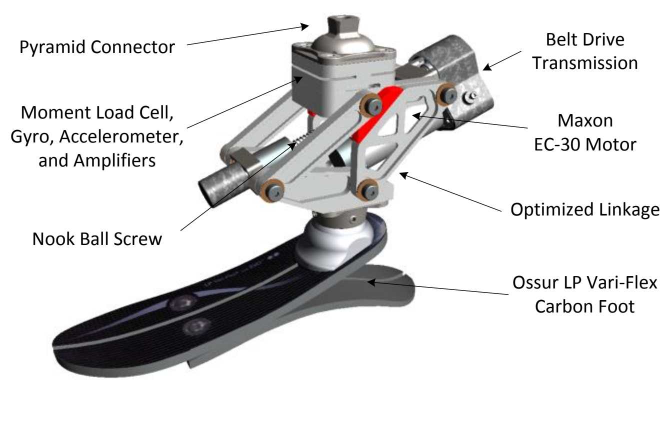

Figure 2. Dynamic Joint Alignment prosthesis uses a four-bar linkage with unequal link lengths that both rotate and translate the joint when actuated.

seen in Fig. 2, has a neutral aligned foot during heel strike, and actively realigns the ankle joint center anteriorly as the foot rotates in a coupled optimized motion. The concept leverages the benefits of different alignments throughout the gait cycle that reduce moments at heel strike and late stance, combined with a fully active prosthesis capable of injecting energy into the stride. A preliminary biomechanics study showed that if the tibia trajectory is altered throughout stance, the peak socket moments can be reduced by up to 50% during late stance [22].

This paper presents a simulation comparison of amputee gait utilizing: a passive carbon spring prosthesis, a powered rotational prosthesis, and the novel design described herein. Sagittal socket moments, joint trajectories and moments, and the trajectory of the pelvis are examined.

II. Modeling and Simulation

Gait simulations were performed using the OpenSim simulation platform [23] to simulate and analyze the stance phase of gait of an able-body model, and three modified models seen in Fig. 3. The OpenSim musculoskeletal model used and modified was 'gait2354' [24]–[26]. Forward dynamics simulations were performed by tracking experimental gait data recorded in the Biomechanics Lab at the University of Massachusetts Amherst and was approved the Institutional Review Board.

For the able-bodied model, segments representing the torso, pelvis, thigh, shank, talus, calcaneus and mid-foot, and toes were scaled in both size and weight to match data collected in the biomechanics lab. The subject that the model scaled to was an able-bodied adult male (27 yr, 70 kg, 1.67 m). The model was limited to 11 degrees of motion similar to what was done in [27], with lumbar flexion added for better control of center of mass without affecting hip joint trajectories. The pelvis had two translational and one rotation degrees of freedom, and all joints were limited to 1 degree of freedom rotational motion to constrain the model to sagittal planar motion. Since the main focus was examining the moment transferred through the socket interface, all joints were actuated with ideal torque actuators constrained to physiological limits based on peak muscle forces.

The modified models represented a transtibial amputee with amputation site mid-tibia to represent a typical transtibial amputee. Tibia mass and inertial properties were modified appropriately. The amputee model was fitted with a passive prosthesis, a pure ankle rotation active prosthesis, and the experimental prosthesis with DJA. The connection to the tibia was a rotational joint located halfway between the amputation site and knee joint to represent an estimated center of rotation of the socket joint in the sagittal plane. Translational movements and rotations in other planes were omitted for model simplicity. A high stiffness of 10,000 N-m/deg and damping of 1000 N-m-s/deg was applied to the joint to represent a rigid ideal socket connection [20]. The passive prosthesis was modeled as a rigid socket, shank, and foot. The pure rotation active prosthesis is modeled as a rigid socket and pylon connected to the actuator assembly with a weld joint. The actuator assembly is connected to the foot with a one degree-of-freedom joint allowing only for sagittal rotation. The power of the pure rotation active prosthesis is the same as an able-bodied human ankle. The DJA prosthesis is modeled as a rigid socket and pylon connected to the linkage assembly consisting of four links and four joints in a closed chain permitting a single coupled motion consisting of one rotational and two translational components. The coordinated movement is determined by the link lengths and attachment angles to the pylon and foot [22]. All of the prostheses masses and inertial properties were calculated using PTC Creo Parametric CAD software.

A simulation of each model was performed tracking the stance phase of gait from heel strike to toe off of able-bodied biomechanics matching the scaled model. The simulations implemented the Reduced Residual Algorithm (RRA) which is a forward integration tool that applies minimized residual forces to the pelvis compensating for inaccuracies in the model representing the subject and actual biomechanics recorded. In this study recorded ground reaction forces were

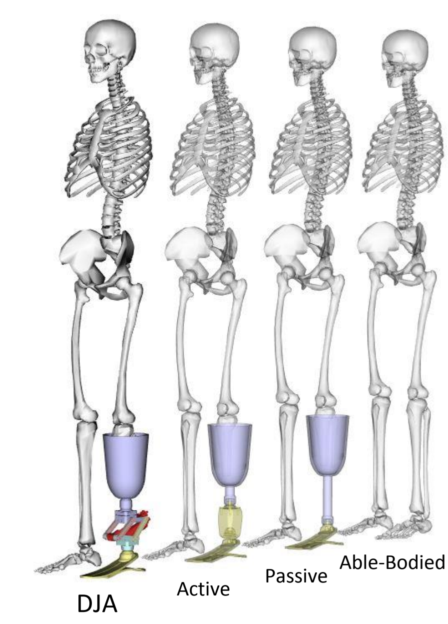

Figure 3. Simulation models used for comparison of biomechanics and socket moments.

prescribed to the model rigid feet. Affected knee angle is weighted more heavily to examine the effects on the center of mass height.

III. Results

Figure 4 presents the simulation results for the knee angle, knee moment, and pelvis height of all four simulations, and socket moments of the amputee models. For the DJA model, the knee is more extended during mid-stance. The knee trajectories of the passive and pure rotation active replicate the able-bodied knee trajectories. Knee moments in all four models are close to able-bodied values. The pelvis height trajectories show that the pure rotation active prosthesis and the DJA prosthesis come close to matching the able-bodied trajectory while the passive prosthesis model clearly drops in late stance due the lack of actuation at the ankle. The pure rotation active prosthesis model produces the highest socket moment, the passive model is slightly lower, and the model with DJA resulted in about 50% lower peak socket moment.

IV. Discussion

The results show a substantial decrease in socket moment in the model that utilizes DJA. This suggests that the maximum pressures in the socket would also be reduced, assuming that pressure on the residual limb is related to the socket moment. The knee moments in all three modified models replicate the able-bodied data. The pelvis height doesn't show a large change for the pure rotation active or DJA prostheses which gives insight into the body's center of mass trajectory and overall efficiency of gait. Noteworthy for the DJA is that the device reduces the socket moment in addition to reducing fluctuations at the pelvis. This suggests that non-physiologically constrained prosthesis designs can be used to restore amputee gait without overburdening the stump-socket interface.

These initial results are limited to the assumptions and simplifications used in the models. In actual amputee gait, the affected joints have altered trajectories to compensate for the deficiencies of the socket-prosthesis system they are using. In these simulation results, the models were programmed to track able-bodied biomechanics. In our approach it does not account for the user's response to loading and the alterations to their gait that they would make to reduce the socket moments. Likewise, the ground reaction forces would most likely be different with the passive prosthesis due to the lack of power and motion. It is assumed that the pure rotation active prosthesis should replicate able-bodied ground reaction forces, for power and motion to be fully restored to the amputee. However, this is not seen in actual amputee gait when a pure rotation active device is used [19]. This may be due to high pressures and a comfort threshold that the amputee may regulate to. Amputee model knee moments are very similar to able-bodied data which would be expected to change with altered kinematics. This may be due to the constraints on the hip angle as well the prescribed GRF which will be addressed in upcoming work for this project.

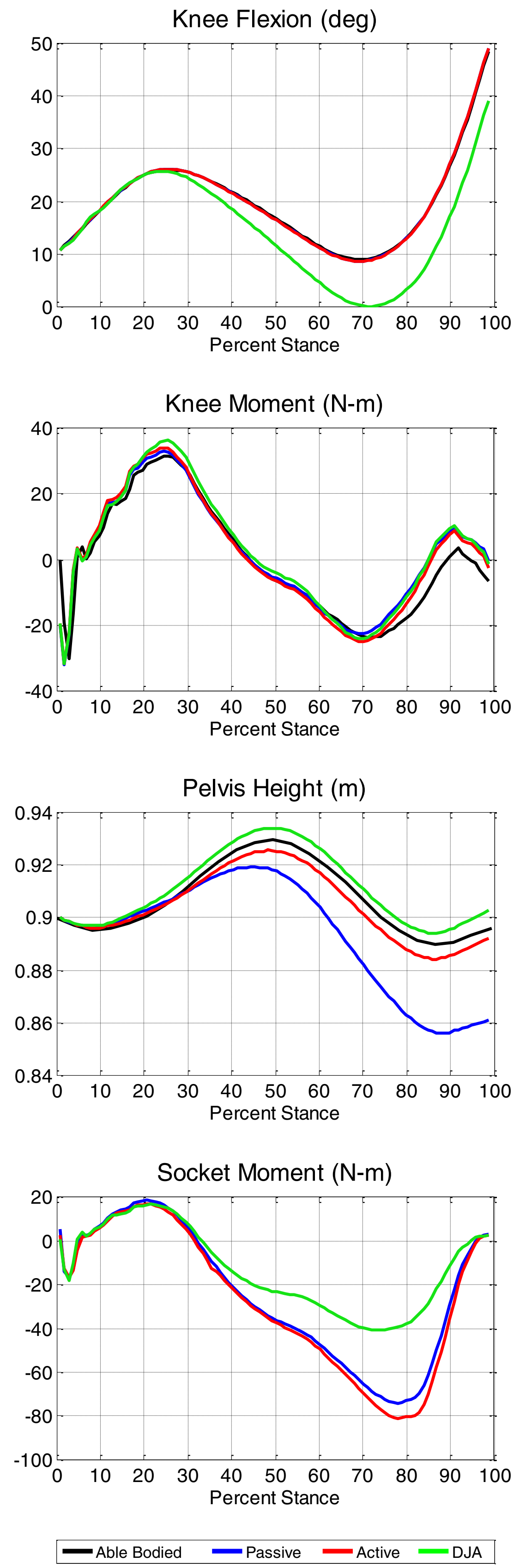

Figure 4. Knee angle, knee moment, pelvis height, and socket moment during stance.

Future simulation work to optimize the design of the DJA device will involve extending the modified models with a multi-segmented, visco-elastic foot (as in [27]) and a foot-ground contact model in order to dynamically predict alterations in the ground reaction force. A more complex model of the stump-socket interface being developed will also be integrated. The models will then be used in predictive forward dynamics simulations minimizing metabolic cost as in [28] and joint loading (as in [29]) to refine the DJA prosthesis design. The prosthesis with DJA will then be tested in the gait lab to validate simulations, and evaluate the performance of the DJA prosthesis. This will lead to a second generation DJA prosthesis design based on both quantitative feedback of gait dynamics and efficiency, as well as qualitative feedback based on the comfort and stability perceived by test subjects.

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