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A Running Controller for a Powered Transfemoral Prosthesis

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Abstract—**This paper describes a running controller for a powered knee and ankle prosthesis. The running controller was implemented on a powered prosthesis prototype and evaluated by a transfemoral amputee subject running on a treadmill at a speed of 2.25 m/s (5.0 mph). The ability of the prosthesis and controller to provide the salient features of a running gait was assessed by comparing the kinematics of running provided by the powered prosthesis to the averaged kinematics of five healthy subjects running at the same speed. This comparison indicates that the powered prosthesis and running controller are able to provide essential features of a healthy running gait.**

I. Introduction

In 2005, approximately 623,000 cases of lower limb amputation existed in the United States, with the total number of cases of limb loss expected to increase by approximately 40% by the year 2020 [1]. Lower-limb prostheses exist in large part to improve the mobility of the user, particularly concerning activities involved in daily living. Dedicated sports prostheses also exist for activities such as running, cycling, swimming, golf, field events, etc., which are used for competition and recreation alike [2, 3]. Many of these devices have been proven quite effective, some having even been accused of providing an unfair advantage to the user [3]. However, should the need arise during the course of normal daily activity for a lower-limb amputee to run—perhaps to quickly dodge an oncoming vehicle or to catch a bus—the individual's daily use prosthesis would be called upon to meet that need.

The majority of prostheses currently available to lower limb amputees are energetically passive. A study of the walking and running gait of paralympic transfemoral amputees (using passive prostheses) revealed gait asymmetry in walking which increased significantly with running; a shortened step length in running was also reported in [4]. Passive prostheses are unable to reproduce the biomechanics of healthy running due in part to the significant net positive power requisite at both the knee and ankle joints [5, 6].

In recent years, prostheses which are able to produce net positive power at the knee or ankle joints have started to emerge [7-10]. However, none of these devices incorporate both a powered knee and ankle; moreover, to the knowledge of the authors, none of these devices have demonstrated restoration of healthy gait characteristics for running in transfemoral amputees. Relative to walking, biomechanically healthy running is characterized by a substantially greater degree of stance knee flexion and a correspondingly greater degree of stance ankle dorsiflexion. Further, the stance phase of running constitutes less than 50% of the stride cycle, while the stance phase of walking constitutes greater than 50% [5, 11]. As such, a walking gait is typically characterized by a double support phase, while a running gait is typically characterized by a double float phase.

This paper presents a control algorithm that enables a running gait in transfemoral amputees. The control algorithm was implemented in a powered knee and ankle prosthesis and tested by a transfemoral amputee running on a treadmill at 2.25 m/s (5.0 mph). The data presented indicate that the controller and prosthesis enable a running gait closely representative of biomechanically healthy running, including the appropriate aforementioned joint kinematics and the double float phase of gait.

II. Controller Design

The controller of the powered prosthesis is structured in three levels. The lowest level controls torque at both the knee and ankle joints. The torque references for each of the joints are generated by a middle level controller, which is implemented as a finite-state machine where each state is defined by passive impedance characteristics for both the knee and ankle. Specifically, the required (knee and ankle) joint torques in each state are characterized by a set of impedance parameters corresponding to the following model

$$\tau = k_i(\theta - \theta_{ei}) + b_i\dot{\theta} \qquad (1)$$

where k_i, b_i, and θ_{ei} denote linear stiffness, damping coefficient, and equilibrium angle, respectively, for the i^{th} state during a gait cycle. Transitions between gait modes or states are triggered by certain biomechanical conditions being met. A separate middle level controller exists for each activity implemented in the prosthesis; at any given time during operation the appropriate middle level controller is selected by the highest level (supervisory) controller. A recent implementation of the supervisory-level controller is described in [12].

In this work, a finite-state (middle level) controller was developed for running gait; the supervisory controller was disabled and the prosthesis restricted to operate only in the running activity mode. Gait modes were determined by an iterative least squares regression application of (1) to a set of running gait data [5] in order to specify the smallest number of (stable) gait modes which sufficiently modeled healthy running. A model with five distinct gait modes and sets of parameters resulted. As this regression did not account for

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dynamic differences between the native limb and the powered prosthesis, the resultant parameters served as a base from which the appropriate parameters were manually tuned.

Within the running controller, these states or gait modes were defined as landing (mode 0), push-off (mode 1), toe-off (mode 2), swing flexion (mode 3), and swing extension (mode 4). These modes and their transition conditions are depicted in Fig.1.

Modes 0 - 2 are stance modes, which in healthy running biomechanics should comprise less than 50% of a stride; modes 3 and 4 are swing modes, which in healthy running biomechanics comprise greater than 50% of a stride [5, 11]. While in mode 0, both the knee and ankle have a relatively high stiffness. The knee flexes in a controlled manner, providing shock absorption and bearing the user's weight. The ankle initially plantarflexes in order to reach a flat-foot state and then dorsiflexes as the user's body center passes over the foot. During mode 1, the knee and ankle actively extend and plantarflex, respectively, in order to propel the user forward and upward. During mode 2, the knee begins to flex as the ankle continues to plantarflex, which assists in flexion of the knee. During mode 3, the knee flexes, and the ankle returns to a slightly dorsiflexed state in order to prepare for the next heel strike. During mode 4, the knee extends, preparing for heel strike. Although not explicitly shown in the state flow diagram in Fig. 1, if during any aerial mode (modes 3 and 4) a load is detected, the controller immediately transitions to the landing mode (mode 0).

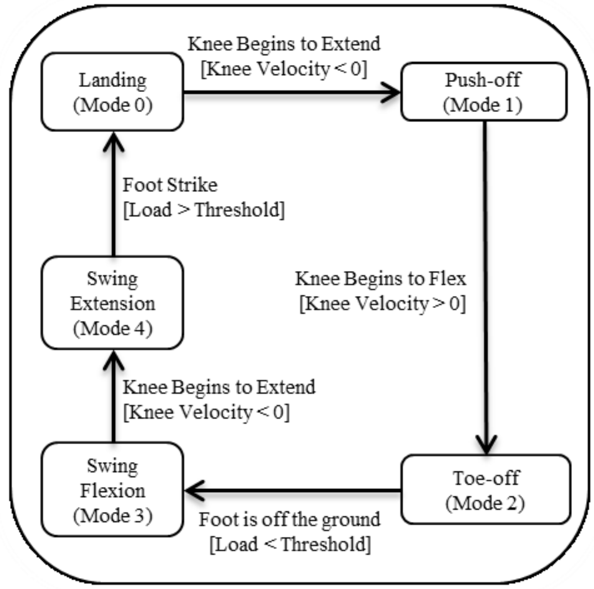


Figure 1. Finite-state model of running gait. Each box represents a state; the respective state transitions are indicated.

III. CONTROLLER DEVELOPMENT

The controller's basic function was verified by a healthy subject fitted with an able-bodied adapter similar to the one described in [13], immobilizing the user's knee at roughly 100 degrees of knee flexion. Once this preliminary verification was complete, the prosthesis was fitted to a unilateral transfemoral amputee, and the impedance parameters were tuned to suit the gait biomechanics of the amputee subject. Approval for the studies described in this work was obtained from the Vanderbilt University Institutional Review Board.

A. Evaluation Metrics

The overarching goal of this work is to enable or improve running gait in the user, specifically in situations when it is not feasible for the user to doff his or her daily use prosthesis and don a running prosthesis. It is assumed in this work that the performance objective of the prosthesis is to reproduce, as faithfully as possible, the function provided by an intact limb. Thus, the controller is evaluated based on its ability to provide sagittal plane joint angles representative of healthy running, on the presence of a double float phase [5, 11], and on the degree of consistency in stride-to-stride gait mode transitions.

In order to obtain reference data representative of healthy running, motion capture data was collected on a small set of healthy subjects. For the motion capture study, five healthy subjects—males ages 24-26—each ran on a treadmill at a speed of 2.25 m/s for two trials, forty-five seconds each. Each of the healthy subjects gave informed consent. The motion capture was achieved with twelve OptiTrack S250e high speed infrared cameras running at 120 Hz using ARENA motion capture software. Thirty-four reflective markers were placed on each subject corresponding to a full skeleton (similar to the Helen Hayes marker set). The software's skeleton solver was used to track the subject's motion. The data collected in ARENA was subsequently processed in MATLAB in order to extract lower limb sagittal joint angles. The joint angles were parsed into single strides (twenty strides per trial) and normalized to a time base of 100%. An offset was applied to the ankle for each subject based upon the angle of the foot with respect to the ground plane during a period where the subject's foot was known to be flat on the ground. The mean and standard deviation over all strides were calculated for each joint.

B. Experimental Tuning

The amputee subject who participated in the running controller evaluation was a 23-year-old male, 6 years post-amputation. The subject's amputation was the result of a traumatic injury; his daily use prosthesis was an Otto-Bock C-Leg with a Freedom Innovations Renegade foot. The subject gave informed consent, including permission for publication of photographs.

The middle level running controller impedance parameters were tuned experimentally on the treadmill during two sessions. The impedance parameters and mode transition thresholds employed during the initial controller verification with a healthy subject were used as a starting point for tuning with the amputee subject, though the spring constants in the stance modes were reduced for user comfort. The impedance parameters and transition conditions were iteratively tuned based upon a combination of knee and ankle joint angle data, qualitative video analysis, and user feedback/comfort to produce appropriate kinematics as well as natural gait mode transitions. The experimentally tuned impedance parameters are shown in Table 1.

TABLE I. IMPEDANCE PARAMETERS

Gait Mode	Knee			Ankle		
	k $\left(\frac{\text{Nm}}{\text{deg}}\right)$	b $\left(\frac{\text{Nms}}{\text{deg}}\right)$	θ_e (deg)	k $\left(\frac{\text{Nm}}{\text{deg}}\right)$	b $\left(\frac{\text{Nms}}{\text{deg}}\right)$	θ_e (deg)
0	4.0	0.1	20.0	5.5	0.2	10.0
1	4.5	0.1	23.0	3.0	0.1	-18.0
2	3.5	0.2	70.0	2.0	0.1	-18.0
3	3.5	0.15	70.0	1.0	0.1	5.0
4	0.9	0.15	20.0	3.0	0.2	5.0

IV. EVALUATION

Following tuning, the controller was evaluated in trials in which the amputee subject ran on a treadmill at 2.25 m/s (5.0 mph). The subject was allowed to utilize the treadmill's handrails. Note that, for the amputee subject wearing the powered prosthesis with running controller, this treadmill speed corresponded to a cadence of 130 steps per minute. Fig. 2 depicts six key elements of a stride captured from a video taken during one trial. Fig. 3 depicts the mode transitions (percent of stride) ± one standard deviation as recorded during the running controller evaluations. This figure demonstrates the consistency of gait mode transitions within the running controller. One should note that mode 2 (toe-off) comprises, on average, less than 3% of stride; this mode serves as an overlap for mode 1 in the ankle and mode 3 in the knee, allowing the knee to flex while the ankle continues to plantarflex.

Fig. 4 compares thirteen strides of the amputee subject running on the powered prosthesis to the sagittal plane knee and ankle joint angles of healthy subjects (obtained from the aforementioned healthy subject motion capture study of running, also at 2.25 m/s). One should first note that the standard deviation of the mean for healthy subjects reflects variety in the running gaits of healthy subjects. While the healthy subjects did exhibit overall uniformity concerning the features of the joint angle curves (except in the ankle near toe-off), range of motion varied considerably between subjects. Concerning the powered prosthesis, the key features of the running gait, in both the knee and ankle

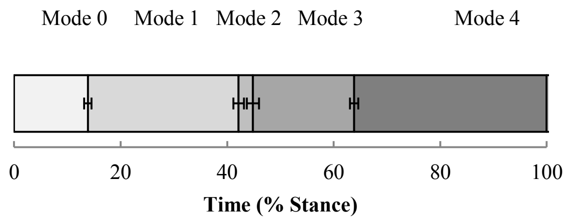
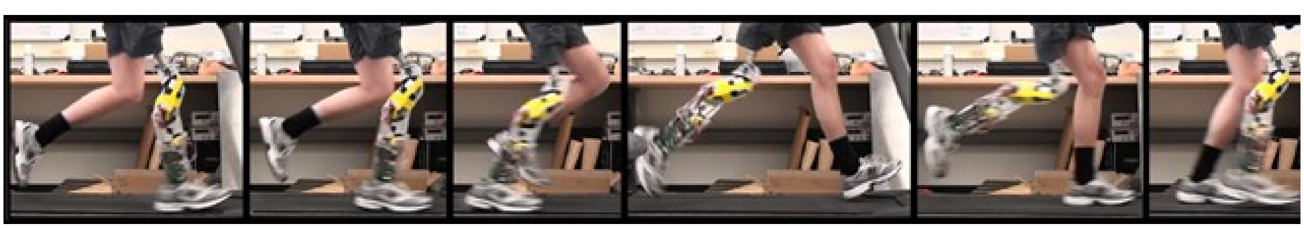


Figure 3. Gait mode transition times with error bars representing one standard deviation

angles, generally match those of the healthy subjects. That is, relative to walking, the knee and ankle joints both achieve a considerably greater degree of flexion and dorsiflexion, respectively, during the stance phase. The two most noticeable deviations between the healthy subject data and the powered prosthesis data are the slight mismatch in kinematics in the mode transition from 0 to 1, and in the decreased swing knee flexion of the prosthesis (i.e., approximately 80 degrees of swing phase knee flexion in the prosthesis, relative to approximately 95 degrees on average in the healthy subject data). The former may indicate the need for an additional gait mode which might better transition between 0 and 1. The latter could be addressed by adjusting the impedance parameters in swing phase. The preference of the amputee, however, may be for somewhat less than a biomechanically healthy amount of swing knee flexion, since the amputee user has limited proprioceptive information from the prosthetic leg, and less swing knee flexion may be more reassuring to the amputee user.

As previously mentioned, another significant feature of running gait is that the stance phase of the latter lasts less than 50% of the stride, which generates a double float phase of gait (as opposed to the double support phase that characterizes walking). Specifically, the stance phase of running has been reported to last between 39% and 45% of the stride [5, 11]. Mode 3, which indicates toe-off in the powered prosthesis gait cycle (i.e., the termination of the stance phase), begins on average at approximately 45% of stride. As such, in this running mode the powered prosthesis provides the relative durations of stance and swing phases that characterize a running gait and distinguish it from a walking gait. Visual evidence of the double float phase of gait, as provided by the powered prosthesis, is shown in Fig. 2.



Heel Strike Landing Push-Off Double Float Swing Flexion Swing Extension

Figure 2. Photo sequence of prosthesis running stride.

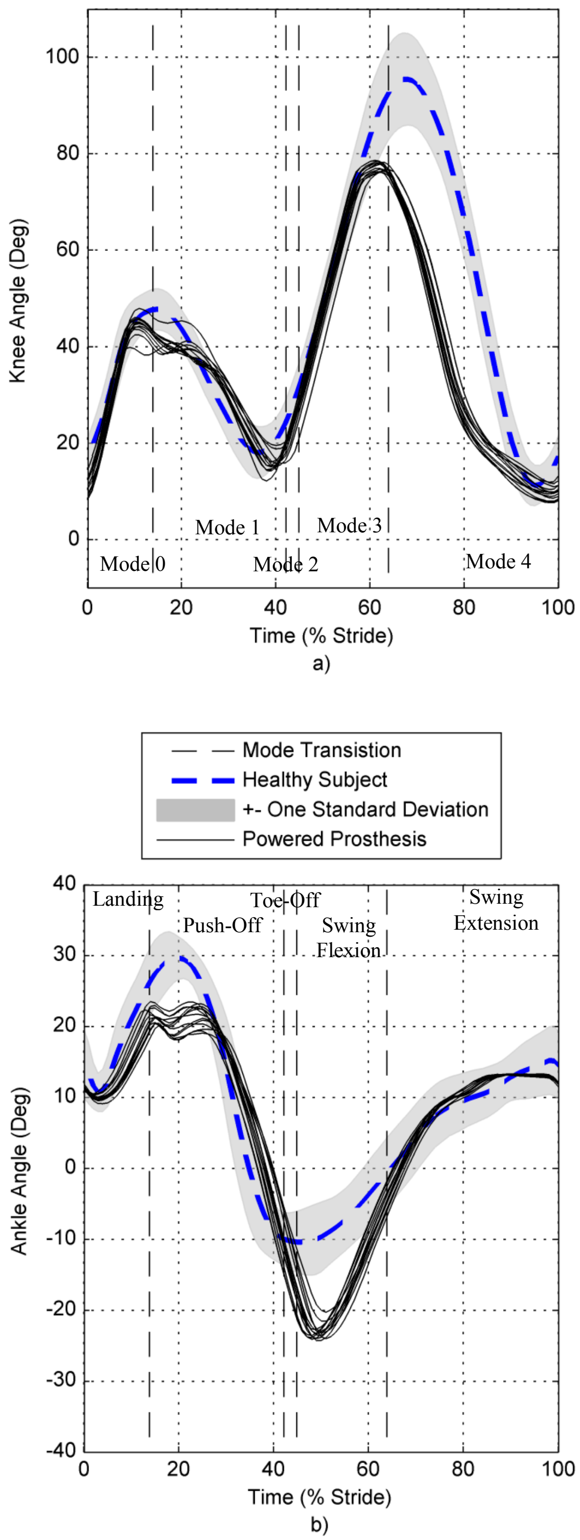


Figure 4. Gait mode transitions, mean joint angles of healthy subjects, and measured joint angle (in degrees) of the powered prosthesis for a) the knee and b) the ankle for 13 strides. Positive joint angle corresponds to dorsiflexion in the ankle and extension in the knee.

V. CONCLUSION

This work presents a running controller implemented in a powered knee and ankle prosthesis. Preliminary evaluation of the controller demonstrated definitive reproduction of a running gait on a treadmill. Future work includes evaluation of running at varying speeds, evaluation of overground running, and automated switching between walking and running gaits.

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