



(51) International Patent Classification:

A61F 2/76 (2006.01) A61F 2/60 (2006.01)
A61F 2/50 (2006.01) A61F 2/68 (2006.01)

(21) International Application Number:

PCT/US2024/035847

(22) International Filing Date:

27 June 2024 (27.06.2024)

(25) Filing Language:

English

(26) Publication Language:

English

(30) Priority Data:

63/523,548 27 June 2023 (27.06.2023) US
63/529,917 31 July 2023 (31.07.2023) US

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(81) Designated States (unless otherwise indicated, for every kind of national protection available): AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CV, CZ, DE, DJ, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IQ, IR, IS, IT, JM, JO, JP, KE, KG, KH, KN, KP, KR, KW, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, MG, MK, MN, MU, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, ST, SV, SY, TH,

(54) Title: POWERED PROSTHESIS WITH IMPROVED SIT-STAND TRANSITIONS

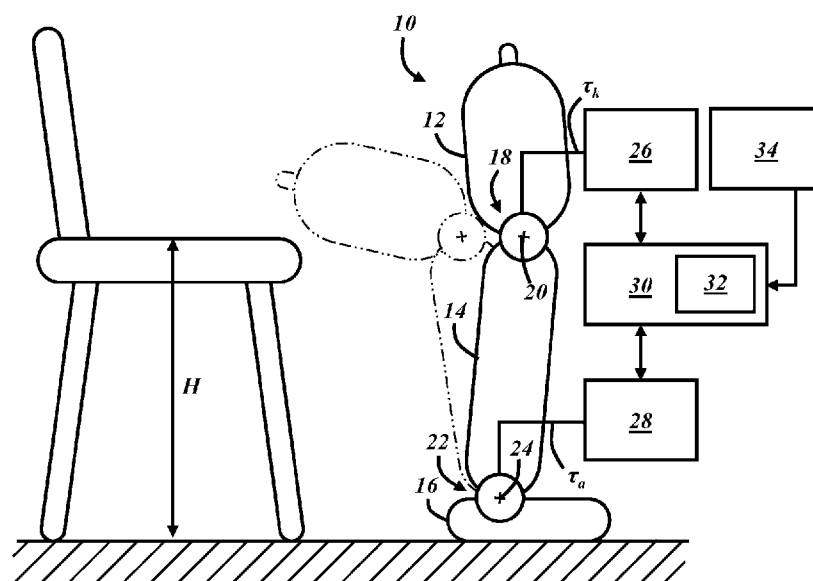


FIG. 2

(57) **Abstract:** A powered knee-ankle prosthesis includes an impedance controller to improve transitions between standing and sitting positions and to permit use at various chair heights. A thigh-based phase variable is used to parameterize optimized data-driven impedance parameter trajectories for sitting, standing, and walking, with only two classification modes. Stand-to-sit and sit-to-stand equilibrium angles are decoupled via a knee velocity-dependent scaling term, which significantly reduces model-fitting error. The prosthesis produces biomimetic joint mechanics, resulting in improved loading symmetry and reduced time to complete sit-to-stand tasks compared to passive prostheses. The controller can be integrated with an impedance-based walking controller to facilitate sit-to-walk transitions from different chair heights. The controller's biomimetic assistance can reduce user overreliance on their natural limb, thereby helping improve mobility for people with above-knee amputations.



TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, WS,
ZA, ZM, ZW.

(84) Designated States (*unless otherwise indicated, for every kind of regional protection available*): ARIPO (BW, CV, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SC, SD, SL, ST, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, ME, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Declarations under Rule 4.17:

- *as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))*
- *as to the applicant's entitlement to claim the priority of the earlier application (Rule 4.17(iii))*

Published:

- *with international search report (Art. 21(3))*
- *before the expiration of the time limit for amending the claims and to be republished in the event of receipt of amendments (Rule 48.2(h))*

POWERED PROSTHESIS WITH IMPROVED SIT-STAND TRANSITIONSGOVERNMENT LICENSE RIGHTS

This invention was made with government support under HD094772 awarded by the National Institutes of Health. The government has certain rights in the invention.

TECHNICAL FIELD

This disclosure is related to powered prostheses and control strategies intended to make artificial joint movement more natural.

BACKGROUND

Conventional passive and semi-active lower leg prostheses cannot supply net positive energy like biological joints. This causes users to compensate with their natural limb, which can lead to kinematic and kinetic asymmetries during walking and during transitions between sitting and standing. Such compensations can cause secondary complications such as osteoarthritis and lower back pain. Powered prostheses have the ability to supply net positive energy and can help reduce secondary complications by producing more normative mechanics. However, effective prosthetic control strategies, particularly for non-rhythmic tasks, remain elusive. The majority of research and development on controllers for powered knee-ankle prostheses has focused on control strategies for rhythmic locomotion—i.e., steady-state walking on level ground. But the reality is that almost half of a person's movement bouts last less than 12 steps each, and a healthy adult transitions between sitting and standing more than 60 times each day on average. For powered prostheses to ever become clinically viable, control strategies accounting for transitional activities will be necessary.

Generally, joint control strategies in powered prostheses include kinematic control strategies and impedance control strategies. Kinematic strategies seek to control joint angles in a manner that replicates healthy human motion. Impedance control, on the other hand, seeks to control resistance to movement about joints by treating each joint as a spring and damper system and varying the spring constant and damping coefficient via application of torque at the joint in a manner that simulates human musculoskeletal resistance to motion. Impedance control dictates joint torque as a function of the joint's angular position θ and velocity $\dot{\theta}$, parameterized by a stiffness K , damping coefficient B , and equilibrium angle θ_{eq} :

$$\tau = K(\theta_{eq} - \theta) - B\dot{\theta}. \quad (1)$$

The small number of attempts made using impedance control for sit-stand motions typically divide those motions into discrete segments and assign constant values to K , B , and θ_{eq} for each segment, using separate controllers for stand-to-sit and sit-to-stand movements and tuning the multiple sets of constants for each individual user. Attempts at kinematic controllers have either failed to provide enough torque to mimic human knee strength or have provided sufficient torque at the expense of asymmetry at critical portions of the motion. In all cases, the time and effort required for tuning the prosthetic to the individual for multiple discrete segments of the sit-stand movements is excessive, requiring up to five hours with certain multi-activity controllers.

In addition to the practical challenges of long tuning times, the risk of task misclassification increases with the number of distinct controllers employed for different activity modes. Such misclassifications can cause unwanted prosthesis behavior ranging from mildly uncomfortable to highly dangerous and likely to result in a fall, depending on the type and timing of the misclassification. Some studies have presented strategies in which transitions from one task to another among sitting, standing, and walking were properly identified but included a percentage of false positives. Other studies have used measured electromyography (EMG) from the intact biceps femoris as an input to a controller that allows for transitions between different tasks with a single controller. However, EMG as a source for a real-time input signal is limited by its tendency to drift over time, which leads to the need for frequent recalibration.

SUMMARY

An embodiment of a powered prosthesis include a joint and an impedance controller configured to continuously vary impedance at the joint during a user transition between a standing position and a seated position.

Another embodiment of the powered prosthesis includes all of the features of the previously listed embodiment, and the impedance at the joint is a continuous function of one or more impedance parameters, each impedance parameter being a continuous function of a sit-stand phase defined between the standing position and the seated position.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and each impedance parameter function is optimized based at least in part on able-bodied data independent from the user.

5 Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the controller estimates the sit-stand phase in real-time.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the controller estimates the sit-stand phase based at least in part on a phase variable that monotonically increases or decreases during the user transition.

10 Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the phase variable is a thigh angle of the user.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and at least one of the one or more impedance parameters is also a function of a decoupling variable which is zero when the user transition is from the standing position to the seated position and non-zero when the user transition is from the seated position to the standing position.

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Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and an equilibrium angle of the joint is one of the impedance parameters and the decoupling variable is an angular velocity of the joint.

20 Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the joint is a knee joint.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the impedance controller is configured to continuously vary impedance at the joint both during a user transition from the standing position to the seated position and during a user transition from the seated position to the standing position.

25 Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the joint is a knee joint and the prosthesis further comprises an ankle joint, the controller being configured to continuously vary impedance at both joints during the user transition.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and further includes a hybrid walking controller configured to control impedance at both joints during a stance phase of walking and to control kinematics of both joints during a gate phase of walking, wherein both controllers control impedance as a function of a thigh angle of the user.

Another embodiment of the powered prosthesis includes all of the features of any of the previously listed embodiments, and the prosthesis is configured to transition between a sit-stand mode and a walk mode based at least in part on a thigh angle and a thigh angular velocity of the user.

BRIEF DESCRIPTION OF DRAWINGS

FIG. 1 is a block diagram of a hybrid kinematic impedance controller (HKIC);

FIG. 2 is a schematic side view of an illustrative powered knee-ankle prosthesis at end phases of a stand-to-sit cycle and a sit-to-stand cycle;

FIG. 3 schematically illustrates criteria for determining occurrence of a stand-to-walk transition and a walk-to-stand transition;

FIG. 4 illustrates optimized impedance parameter functions for the knee and ankle joints during sit-to-stand and stand-to-sit transitions based on able-bodied datasets;

FIG. 5 illustrates average phase, kinematic, and kinetic trajectories observed during sit-to-stand and stand-to-sit trials with a powered knee-ankle prosthesis equipped with an impedance controller in comparison to able-bodied data;

FIG. 6 illustrates differences in degree of asymmetry (DoA) and RMS loading between powered (impedance-controlled) and passive prostheses during sit-to-stand and stand-to-sit trials at multiple chair heights;

FIG. 7 illustrates differences in degree of asymmetry (DoA) between powered (impedance controlled) and passive prostheses during rapid sit-to-stand and stand-to-sit trials at multiple chair heights; and

FIG. 8 illustrates differences in completion times between powered (impedance controlled) and passive prostheses during trials including a sit-stand-walk transition and a walk-stand-sit transition.

DESCRIPTION OF EMBODIMENTS

Described below is a continuous control framework for a powered knee-ankle prosthesis that enables sitting, standing, and walking, as well as transitions between them. A unified data-driven impedance control model for sitting and standing movements is presented that can be integrated with a hybrid walking mode that employs both impedance control and kinematic control. The sit-stand control model uses an equilibrium angle θ_{eq} that is dependent on knee velocity, which facilitates distinct behaviors for sit-to-stand and stand-to-sit tasks without explicit task classification. Autonomous phase variable detection is employed for sitting and standing movements to allow the controller to adapt to different chair heights and standing postures. In trials with above-knee amputee users with multiple chair heights, the sit-stand control strategy demonstrates improved similarity to able-bodied mechanics when compared to previous work, as well as improved clinical outcome metrics such as loading symmetry and time to complete a given task without the use of EMG. When integrated with a hybrid walking controller, this sit-stand impedance control model has demonstrated efficacy in more realistic scenarios, including transitions from sitting at different chair heights to walking.

Commonly assigned international patent application number PCT/US2024/016734 by Gregg, et al., filed February 21, 2024, is hereby incorporated by reference in its entirety. That application discloses a hybrid kinematic/impedance controller for walking, with continuously variable impedance control employed during the stance phase and kinematic control employed during the swing phase. The impedance controller uses impedance parameters that are continuous functions of gait phase, walking speed, and incline using real-time estimates of walking speed and incline to enable autonomous task adaption. That controller uses global thigh angle θ_{th} as a task-invariant phase variable, estimates gait phase, walking speed, and incline in real time, and is referred to below as the “hybrid kinematic impedance controller,” “HKIC,” or “hybrid walking controller.” A block diagram of the hybrid walking controller is provided in FIG. 1, where real-time estimates of gait phase $\hat{s}$ and conditions $\hat{\chi}$ —i.e., walking speed and incline—define the joint impedance parameters K , B , θ_{eq} and joint angles θ_d using data-driven models. Depending on whether the user and prosthesis are in stance or swing phase, torque commands τ are calculated using either an impedance controller or a kinematic controller, respectively. Once configured with a user’s mass and leg segment lengths, the hybrid

walking controller can operate autonomously, requiring neither manual impedance tuning nor external knowledge of the terrain.

The sit-stand impedance controller discussed below can be integrated with the hybrid walking controller to enable improved sit-stand-walk and walk-stand-sit transitions. Similar to the HKIC, the sit-stand controlled relies on a novel, data-driven impedance parameter model based on able-bodied data that produces biomimetic joint torque during both sitting and standing motions. To synchronize that model with the user's motion, an adaptive sit-stand phase variable is employed that automatically adjusts to chair height and standing posture.

As used herein, a "variable impedance controller" is a controller that continually varies the output mechanical impedance at a prosthetic joint. Mechanical impedance is generally a measure of resistance to movement from an equilibrium position, which in this context is resistance to rotation about a joint axis. Resistance to rotation about a joint axis is a function of joint stiffness, joint damping, and joint angle relative to an equilibrium angle according to equation (1). Mechanical impedance at a prosthetic joint can be provided via an applied torque in a rotational direction about the joint axis. This torque can be applied by an electric motor coupled with a transmission or other suitable means. Impedance control is different from kinematic control in that impedance is a measure of resistance to movement relative to an equilibrium position in a freely rotatable joint, while kinematic control positively controls joint movement by rotating an object about the joint axis from one angular position to another with a particular angular speed function via a motor and transmission or other suitable means.

FIG. 2 schematically illustrates an example of a powered knee-ankle prosthesis 10 that includes an upper leg member 12, a lower leg member 14, and a foot member 16. The upper leg member 12 is adapted for attachment to the end of the leg of an above-knee amputee user at one end and is coupled with the lower leg member 14 at a knee joint 18. The knee joint 18 is a rotational joint that provides rotational movement of the lower leg member 14 relative to the upper leg member 12 about a knee axis 20. The lower leg member 14 extends from the knee joint 18 to an ankle joint 22 at which the foot member 16 is coupled with the lower member 14. The ankle joint 22 is a rotational joint that provides rotational movement of the foot member 16 relative to the lower leg member 14 about an ankle axis 24.

The prosthesis 10 includes a first actuator 26 (e.g., a motor) configured to provide a knee torque τ_k in a rotational direction at the knee joint 18 and a second actuator 28 configured to provide an ankle torque τ_a in a rotational direction at the ankle joint 22. For example, the first actuator 26 may be rigidly mounted along the upper leg member 12, and its rotational output may be converted to the torque τ_k applied to the knee joint 18 via a transmission member rigidly attached to the lower leg member 14. Similarly, the second actuator 28 may be rigidly mounted along the lower leg member 14, and its rotational output may be converted to the torque τ_a applied to the ankle joint 22 via a transmission member rigidly attached to the foot member 16. Other arrangements are possible to provide torque at one or more of the prosthetic joints.

The prosthesis 10 includes at least one controller 30 configured to store and employ a control scheme 32 according to which the controller operates each actuator 26, 28 to provide the desired torque τ_k, τ_a at each joint 18, 22. The controller 30 is programable and in communication with each actuator 26, 28 to control its output torque. In this example, the controller 30 is in two-way communication with each actuator 26, 28 to receive one or more inputs from the actuators, such as a real-time encoder position which can be used to determine real-time angular velocities at each joint 18, 22, among other parameters. The controller 30 may receive additional information from one or more sensors 34 (e.g., an accelerometer) to implement the control scheme 32.

FIG. 2 schematically illustrates the prosthesis 10 at the two end phases of a stand-to-sit or a sit-to-stand cycle. The prosthesis 10 is illustrated in solid lines at the stand phase, which is at the beginning of a stand-to-sit cycle or at the end of a sit-to-stand cycle. The prosthesis 10 is illustrated in phantom lines at the sit phase, which is at the beginning of a sit-to-stand cycle or at the end of a stand-to-sit cycle. Between the stand phase and sit phase is a transient phase in which the prosthesis is changing and moving from one end phase to the other. Each joint impedance parameter is a continuous function between the end phases. As used herein, a “phase” is a point along a continuous function between 0 and 1, where 0 is the sit phase and 1 is the stand phase. The control scheme 32 may be an impedance control scheme used by the controller 30 to provide torque commands to the actuators 26, 28 during sitting and/or standing movements. When combined with walking control, any combination of controllers and control schemes may be included as part of the prosthesis 10. For example, when combined with the above-described hybrid walking controller, the prosthesis may include a position controller operable during the swing phase of the user’s gait, a first impedance

controller operable during the stance phase of the user's gait, and a second impedance controller operable during a sit or stand task. Or one impedance controller can control joint torque based on more than one control scheme selected based on task identification. In other words, torque control at the joints 18, 22 is not limited to any particular controller or number of controllers. Rather, one or more controllers are used to control joint torque according to each of one or more control schemes. While FIG. 2 illustrates only one controller 30 and associated control scheme 32, some implementations include multiple controllers. The prosthesis 10 may for example include an impedance controller with one or more impedance control schemes and a separate kinematic controller with one or more kinematic control schemes. The prosthesis 10 may include dedicated controllers for each joint as well.

The prosthesis 10 may include a high-level finite state machine (FSM) with two or more modes to determine the user's intent based on prosthesis sensor readings. The modes include a sit-stand mode and a walk mode. The sit-stand mode and walk mode may be the only two modes, with the sit-stand mode being automatically selected when it is determined that the user is standing (i.e., not walking), and the walk mode being automatically selected when it is determined that the user's intent is to walk. Limiting the FSM to only two control modes reduces complexity and the chances of incorrect task or transition identification. However, additional modes are not excluded, as the impedance control schemes disclosed herein can be implemented with any number of prosthesis modes.

The FSM may implement one or more transition rules based on measurements from the one or more sensors 34. These measurements may include global thigh angle θ_t and thigh angular velocity ω_t , obtainable, for example, from an inertial measurement unit (e.g., a MicroStrain 3DM[®] CX5-25 attitude reference sensor, Parker LORD, Williston, VT) mounted to the proximal end of the knee joint. The measurements may also include knee angle θ , obtainable from a rotary encoder (e.g., E5 Series, 3600 CPR, US Digital, Vancouver, WA) at the knee joint 18, and/or the presence of foot-ground contact (FGC), obtainable from a load cell (e.g., M3564F load cell, Sunrise Instruments, Nanning, China) at the distal end of the ankle joint. These or other sensor signals can be used to heuristically determine criteria for identifying a transition to and from walk mode and sit-stand mode.

In one embodiment, and with reference to FIG. 3, the transition rules include two stand-to-walk criteria, as the user U can initiate walking movement from the stand phase by leading with either

the prosthetic 10, indicated by criteria I, or the biological leg, indicated by criteria II. The FSM may thus transition from sit-stand mode to the walk mode either if a prosthesis-side heel strike is detected, per criteria I, or if a prosthesis-side late stance of a gait cycle is detected, per criteria II. Both sets of criteria include characteristic foot-ground contact signals F_{CG} with a first threshold defining the making of a new ground contact and a second threshold defining the breaking of an existing ground contact. The first threshold may be greater than the second threshold. In one embodiment, the first threshold is in a range between 50 N and 100 N, such as 75 N, and the second threshold is in a range between 0 N and 50 N, such as 25 N.

Criterion I includes a characteristic foot-ground contact signal F_{CG} while the global thigh angle θ_t and thigh angular velocity ω_t are within prescribed limits. The contact signal F_{CG} for the heelstrike transition from standing to walking is characterized by a rising edge in the signal. The thigh angle θ_t for the heelstrike transition from standing to walking is in flexion and is in a range with a lower limit greater than 0. One suitable range for thigh angle θ_t is between 10 and 40 degrees in flexion ($10^\circ < \theta_t < 40^\circ$). The thigh angular velocity ω_t for the heelstrike transition from standing to walking is in the extension direction and has an upper limit. The upper limit on thigh angular velocity ω_t for the heelstrike transition from standing to walking may be in a range from 20 to 25 degrees per second, such as 23 degrees per second, in the extension direction ($\omega_t < -23^\circ/\text{sec}$). The upper bound on thigh angle θ_t prevents erroneous transitions to walk mode while the user is seated.

Criterion II includes the presence of a contact signal F_{CG} based on the above-mentioned first and second thresholds—e.g., a signal that has reached 75 N and has not fallen below 25 N—with prescribed upper limits for thigh angle θ_t and thigh velocity ω_t . The upper limit on θ_t for late stance transition to walking may be in a range from 10 to 20 degrees, such as 15 degrees, in extension ($\theta_t < -15^\circ$). The upper limit on thigh angular velocity ω_t for late stance transition to walking may be the same as the upper limit for heelstrike transition—i.e., in a range from 20 to 25 degrees per second, such as 23 degrees per second, in the extension direction ($\omega_t < -23^\circ/\text{sec}$).

With continued reference to FIG. 3, the transition rules may include only one set of walk-to-stand criteria. Specifically, the FSM may transition from walk mode to sit-stand mode when a ground contact signal F_{CG} based on the above-mentioned first and second thresholds is present with θ_t , ω_t , and knee angle θ_k within prescribed limits indicative of upright stance. The absolute value of each of the thigh angle θ_t and thigh velocity ω_t may have upper limits in a range between 5 and 15 degrees.

One suitable set of limits for thigh angle is 10 degrees in flexion or extension ($|\theta_t| < 10^\circ$), and one suitable set of limits for thigh velocity is 11 degrees per second in the flexion or extension directions ($|\omega_t| < 10^\circ/\text{sec}$). To prevent rapid FSM switching that could occur if the user pauses during toe-off, transition from walk mode to sit-stand mode may be disallowed if knee angle θ_k is greater than a prescribed upper limit. This upper limit may be in a range between 10 and 20 degrees, such as 15 degrees ($\theta_k > 15^\circ$).

During each transition from walk mode to sit-stand mode and from sit-stand mode to walk mode, the torque commands sent to the actuators 26, 28 can be based on the control torque prescribed by both modes. For example, the torque command for each joint 18, 22 may be calculated as a convex weighted sum of the respective control torques given by the walk mode and the sit-stand mode. The weight of each respective control torque may vary linearly over a prescribed time period to produce a smooth mode transition with minimal jerk. The time period may be in a range from 100-300 ms (e.g., 200 ms).

Sit-Stand Phase Variable

A sit-stand phase variable can be defined based on the user's prosthetic global thigh angle θ_t to parameterize impedance controller behavior during sit-stand motions. Global thigh angle θ_t is an appropriate choice for a phase variable in this case because it monotonically increases during sit-to-stand motion and monotonically decreases during stand-to-sit motion. During sit-stand movement, the prosthetic 10 has a continuous set of transitional phases s between a sitting state ($s = 0$) and a standing state ($s = 1$). The phase variable s can increase or decrease, making both sit-to-stand and stand-to-sit motions possible with a single controller. Similar to the above-described hybrid walking controller, the phase variable s can be defined through an affine transformation of θ_t :

$$s = (\theta_t^0 - \theta_t) / (\theta_t^0 - \theta_t^f), \quad (2)$$

where θ_t^0 is the value of θ_t when the user sits on a surface, and θ_t^f is the value of θ_t when the user stands comfortably. Typically, θ_t^f is near 0° but may vary if a user prefers standing with one foot more anterior than the other, for example.

The closed chain kinematics of sitting make θ_t^0 depend on the dimensions of the user's leg and on the chair height H (FIG. 2). To account for varying user anatomies and chair heights, θ_t^0 may be calculated in real-time with a moving average (e.g., over a 0.5 sec sliding window) while static

sitting conditions are met. The criteria for a static sitting condition may be based on prescribed limits for thigh angle and velocity θ_t , ω_t . For static sitting, thigh angle may have a lower limit between 40 and 50 degrees, such as 45 degrees in flexion ($\theta_t > 45^\circ$), and thigh velocity may have an upper limit between 5 and 10 degrees per second, such as 6 degrees per second, in flexion and extension ($|\omega_t| < 6^\circ/\text{sec}$). Similarly, θ_t^f may be calculated in real-time with a moving average while static standing conditions are met. The criteria for a static standing condition may be based on prescribed limits for thigh angle, thigh velocity, and F_{CG} . For static standing, thigh angle may have an upper limit between 10 and 20 degrees, such as 15 degrees in flexion ($\theta_t < 15^\circ$), thigh velocity may have an upper limit between 5 and 10 degrees per second, such as 6 degrees per second, in flexion and extension ($|\omega_t| < 6^\circ/\text{sec}$), and ground contact force may have a lower limit between 100 N and 300 N, such as 200 N ($F_{CG} > 200 \text{ N}$).

Sit-Stand Impedance Model

The impedance model may be trained with able-bodied reference data collected during sitting and standing motions. For purposes of this disclosure, the reference data included kinetic and kinematic data collected from eight able-bodied participants during five consecutive sit-to-stand and stand-to-sit transitions, each at a self-selected pace of the respective participant. From these data, sagittal plane kinematics and kinetics for the knee and ankle joints can be used, and θ_t in the sagittal plane can be calculated using hip and torso kinematic data. Joint torques may be normalized by subject mass, and joint angles may be differentiated and filtered (e.g., with a fourth order Butterworth low-pass filter) to obtain angular velocities. While the cut-off frequency often used during walking is 5-6 Hz, a lower cut-off frequency may be used for sit-stand transitions due to their slower frequency content. 2 Hz is one suitable cut-off frequency.

The able-bodied data can then be segmented to determine the start and end of the sit-to-stand and stand-to-sit motions. The start and end of each motion can be defined at least in part by a thigh velocity threshold. The start of a sit-to-stand or stand-to-sit motion can be defined in the able-bodied data where the thigh velocity crosses the threshold before the motion, and the end of the same motion can be defined where the thigh velocity crosses the threshold after the motion. One suitable threshold is at $|\omega_t| < 5^\circ/\text{sec}$. Each able-bodied subject's average minimum knee angle may be subtracted out over all data collection trials. This is in response to some subjects consistently reporting a knee flexion

angle much larger than zero during standing, which is believed to be an artifact of motion capture rather than a true behavior.

From there, a phase variable trajectory for each trial for each able-bodied participant can be calculated using equation (2). The resulting dataset from eight participants performing 5 stand-to-sit-to-stand motions includes 80 sets of knee and ankle kinematic, kinetic, and phase trajectories, half of which are sit-to-stand motions, and half of which are stand-to-sit motions.

Impedance parameters including K , B , and θ_{eq} can be modeled as continuous fourth-order polynomials in phase s :

$$\begin{bmatrix} K(s) \\ B(s) \\ \theta_{eq}(s, \dot{\theta}) \end{bmatrix} = \begin{bmatrix} k^\top \\ b^\top \\ e(\dot{\theta})^\top \end{bmatrix} \begin{bmatrix} s^0 \\ \vdots \\ s^4 \end{bmatrix}, \quad (3)$$

10 where $k \in \mathbb{R}^{4 \times 1}$ and $b \in \mathbb{R}^{4 \times 1}$ are constant coefficient vectors defining the stiffness and damping polynomials, respectively.

The equilibrium angle parameter vector $e \in \mathbb{R}^{4 \times 1}$ can be defined as a function of joint angular velocity $\dot{\theta}$. It has been found that treating the equilibrium angle parameters as constants can result in the model providing excessive knee extension torque at the beginning of the stand-to-sit motion, which can cause the user to struggle to initiate the sitting motion. The added dependency on joint
15 velocity decouples the equilibrium angle trajectories for sit-to-stand and stand-to-sit motions:

$$e(\dot{\theta}) = e_1 + f_j(\dot{\theta})e_2, \quad (4)$$

where $e_1 \in \mathbb{R}^{4 \times 1}$ and $e_2 \in \mathbb{R}^{4 \times 1}$ are constant vectors, and $f_j(\dot{\theta}) \in [0,1]$ is a saturating ReLU function that is active only for the knee joint, defined as:

$$20 \quad f_{knee}(\dot{\theta}) = ReLU\left(\min\left(1, \frac{\dot{\theta}}{\eta}\right)\right), \quad f_{ankle}(\dot{\theta}) = 0. \quad (5)$$

A fixed parameter η controls the width of the linear window. One suitable value for η is 0.5 deg/sec.

The knee kinematic trajectory is monotonic during both sitting and standing motions, meaning that $f_{knee}(\dot{\theta}) = 0.0$ during sit-to-stand (negative knee velocity) and $f_{knee}(\dot{\theta}) \approx 1.0$ during stand-to-sit (positive knee velocity). Therefore, the coefficients defining the equilibrium angle are
25 $e(\dot{\theta}) = e_1$ for sit-to-stand and $e(\dot{\theta}) = e_1 + e_2$ for stand-to-sit. During transitions between either

sitting or standing motions (e.g., $0 < \dot{\theta} < 0.5$ deg/sec), the parameters continuously interpolate between the steady-state cases. In practice, this novel model definition allows for decoupled equilibrium angle trajectories for sitting and for standing, without necessitating discrete task classification between cases.

5 The model can be fully defined through the set of coefficients $\kappa = \{k, b, e_1, e_2\}$. To train the model on the dataset, an optimization problem can be constructed to select the optimal κ such that the impedance control equation (1) best reproduces the normalized torque profiles τ , given the joint kinematics θ , angular velocities $\dot{\theta}$, and phase trajectories s across all trials:

$$\kappa = \arg \min \|\tau - \hat{\tau}\|_2^2, \quad (6)$$

10 where $\hat{\tau} = K(s)(\theta_{eq}(s, \dot{\theta}) - \theta) - B(s)\dot{\theta}$.

The dataset phase trajectories s are included in the optimization, allowing it to internally account for the expected shape of the θ_t trajectory and the resulting nonlinearity in the phase variable. This is an improvement over assuming a perfectly linear phase trajectory, which would necessitate an additional online linearization step.

15 As written, equation (6) is difficult to solve, as the product term $K(s)\theta_{eq}(s, \dot{\theta})$ is nonlinear in the decision variables κ . The approach used in development of the hybrid walking controller in the above-mentioned Gregg application can be used here to approximate the cost function in equation (6) with a convex approximation. Defining $\mathbf{s} = [s^0 \dots s^d]^T$ and $\delta_i(s) = (k^T \mathbf{s})(e_i^T \mathbf{s})$, for $i = \{1, 2\}$, the nonlinear term can be rewritten as $K(s)\theta_{eq}(s, \dot{\theta}) = \delta_1(s) + \delta_2(s)$. By treating the coefficients in
20 the δ_i polynomials as independent decision variables, the cost function becomes linear in the decision variables and reduces to a convex quadratic program. The original coefficients e_1 and e_2 can be recovered from the solution by simply approximating the rational function $\delta_i(s)/K(s)$ with a fourth-order polynomial as in Gregg, et al.

25 Constraints may be added to the optimization problem based on desired controller behavior and pilot testing to ensure reasonable impedance trajectories for both sitting and standing motions. For example, stiffness can be constraint to a prescribed minimum at both the knee joint 18 and the ankle joint 20. Each minimum stiffness may be in a range between 0 and 0.1 Nm/(deg·kg). In the experiments and results discussed below, the stiffness at the knee joint 18 was set to be greater than 0.0087 Nm/(deg·kg), and the stiffness at the ankle joint 22 was set to be greater than 0.0175

Nm/(deg·kg). Damping may also be constrained to a prescribed range between 1×10^{-4} and 3×10^{-3} Nm·sec/(deg·kg), for example. In the experiments and results discussed below, damping was constrained to be between 1.74×10^{-4} and 2.6×10^{-3} Nm·sec/(deg·kg) for both joints.

Soft constraints may also be added in the form of additional cost function terms that penalize constraint violation, for example, to ensure that equation (1) produces no torque at either joint during sitting ($s = 0$) and no torque at the knee joint during standing ($s = 1$). Soft constraints are used to prevent numerical sensitivity, and because small, non-zero joint torques are still acceptable. Given positive stiffness and damping constraints, the sitting and standing torque constraints indirectly constrain $\theta_{eq}(0)$ for both joints and $\theta_{eq}(1)$ for the knee joint to be equal to the mean sitting and standing joint angles, respectively.

FIG. 4 shows the optimized impedance parameter functions for both the knee joint 18 and the ankle joint 22 based on the above-noted able-bodied datasets. Positive equilibrium angles correspond to knee flexion and ankle dorsiflexion. For each parameter, phase progresses from 0 to 1 during sit-to-stand and from 1 to 0 during stand-to-sit. The controller can provide smooth movement between each curve based on equations (4) and (5), providing appropriate assistance levels during both standing and sitting without explicit classification.

The equilibrium angle for the knee joint 18 is illustrated for the both sit-to-stand (positive angular velocity) and stand-to-sit (negative angular velocity) motions in chart (C). As shown in chart (D), optimal ankle stiffness maintains a static value at the minimum constraint. This may be due to the ankle torque in the dataset being fairly small in magnitude and having a large variance. While this suggests that lowering the minimum ankle joint stiffness constraint may increase the fit of the model to the dataset, a controller with a stiffness that is too low would be unable to reject disturbances or robustly handle inter-subject variation.

The quality of the fit of the optimization results to the dataset can be evaluated by calculating the model joint torque using equations (1) and (3) and κ^* at each point in the dataset. The root mean squared error (RMSE) of the model torque can be compared to typical human variation by normalizing it by the standard deviation of the joint torque observed in the dataset.

Experimental

The above-described sit-stand impedance control scheme and controller have been implemented in a powered knee-ankle prosthesis to experimentally evaluate their effectiveness. The powered prosthesis used in these experiments is also equipped with the earlier-described hybrid walking controller. However, it is contemplated that the sit-stand controller detailed in this disclosure could be integrated with other types of walking controllers.

Experiments were conducted with three above-knee amputate participants, and outcomes with the powered prosthesis and sit-stand control scheme were compared to outcomes with the participants' own prescribed prostheses. The demographics of each participant are described below in TABLE I, and the prosthesis prescribed to each participant is listed below in TABLE II.

TABLE I

	Sex	Age	Body Mass	Height	Prosthetic Side	Residual Limb Length	Time Since Amputation	Etiology
			(kg)	(cm)		(cm)	(yrs)	
1	M	26	108.8	192	L	26.8	26	Congenital
2	M	40	79.2	180	L	38	23	Cancer
3	M	18	69.3	183	L	15.9	18	Congenital

TABLE II

	Prescribed Knee-Ankle Prosthesis
1	C Leg 4/Triase
2	Rheo Knee/Freedom Maverick Xtreme AT
3	Genium X3/Pro-Flex LP

The experimental protocol was approved by the Institutional Review Board of the University of Michigan (HUM00166976), and participants wore a ceiling-mounted safety harness for the duration of the experiments. A certified prosthetist assisted in fitting the powered prosthesis, changing out the two prostheses, and ensuring the safety of the participants throughout the experiments.

The experimental protocol investigated sit-to-stand, stand-to-sit, walking, and turning motions while participants wore either their standard prosthesis or the powered prosthesis. Similarity in kinematics and kinetics between the above-described sit-stand impedance controller and able-bodied data were investigated. Also, performance of the powered prosthesis with the above-described sit-stand controller was compared to that of the participants' standard prosthesis in terms of functional metrics. Because passive and semi-active above-knee prostheses are associated with increased asymmetry and completion time in sit-stand transitions compared to able-bodied movement, both of

these metrics were investigated. Using these metrics, both the generalization and functionality of the sit-stand controller were tested with different sitting and standing speeds and chair heights. Finally, the ability of the controller to perform in more “real-world” conditions and with the earlier-described hybrid walking controller was evaluated via a test involving sitting, standing, turning, and walking between chairs of different heights.

The detailed protocol included both training and experimental validation components for both the powered and passive prostheses, and the order in which these two conditions were tested was randomized for the three participants. The powered prosthesis condition included an initial fitting with the help of a certified prosthetist, which included walking between parallel bars to adjust alignment of the prosthetic leg. The sitting and standing movements were performed using a height-adjustable four-legged stool. Two of the participants completed the protocol within a one-day session, and the third participant completed the protocol for each prosthesis on separate days after becoming fatigued on the first day with the first prosthesis condition.

Conditions for both prostheses included approximately 15 minutes of symmetry training, in which the participants practiced sitting and standing with one leg each on two in-ground force plates and from which they received visual feedback regarding the loading symmetry in the vertical direction between the two legs. Of these 15 minutes of training, the first five minutes included instruction on the visual feedback and instructions to try different strategies to see how symmetry was affected. In the next five minutes, the participants were instructed to purposefully sit and stand asymmetrically on one side or another. In the last five minutes, the participants were instructed to try to sit and stand as symmetrically as possible. After the training was complete, the visual feedback was removed.

Following the training, sit-to-stand and stand-to-sit movements were performed with two different speed conditions (self-paced and rapid) and three different stool height conditions (tall, medium, and short, corresponding to 53.5 cm, 51.0 cm, and 48.5 cm, respectively). Participants were asked to keep their hands crossed on their chest during the movements. In the self-paced speed condition, each participant was cued to sit and then stand five times each at a comfortable pace with the goal of maximizing the symmetry that had been practiced with the visual feedback. In the rapid speed condition, each participant was asked to sit and then stand five times each as fast as comfortably possible. Both conditions with five sit-stand cycles were repeated three times with a one-minute break in between, for a total of fifteen sit/stand cycles per condition.

Finally, the participants completed a multi-activity task including a sit-stand-walk sequence including different chair heights. In this test, each participant started seated on the stool at the tallest setting, stood up, walked 12 meters to a chair of height 44.5 cm, sat down on the chair, stood back up, walked back to the starting stool, and sat down on the stool. Each participant was asked to complete this sequence as fast as comfortably possible and repeated the sequence five times with each prosthesis. Participant 1 was only able to complete two trials due to a hardware problem. The duration of each sit-stand-walk sequence was recorded.

The phase, joint angle, and joint torque trajectories observed during the self-paced trials were compared to their equivalents in the able-bodied dataset. Phase was determined using the thigh angle sensor and equation (2), joint angles were measured using joint encoders, and joint torque was calculated using motor current and a validated motor model. Data from the powered prosthesis was segmented in the same manner as the able-bodied dataset in the Sit-Stand Impedance Model section above.

Next, various clinical metrics were calculated and checked for significant differences due to the powered prosthesis. The degree of asymmetry *DoA* in leg loading was calculated for both the self-paced and rapid trials as:

$$DoA = (F_{z,bio} - F_{z,prosth}) / (F_{z,bio} + F_{z,prosth}), \quad (7)$$

where $F_{z,bio}$ and $F_{z,prosth}$ are the vertical forces applied by the biological and prosthetic foot, respectively. To avoid division by zero, datapoints where the combined vertical force from both legs $F_{z,feet} = (F_{z,bio} + F_{z,prosth})$ was less than 10% of the participant's bodyweight were excluded. A positive *DoA* value corresponds to increased loading of the biological limb, and a negative *DoA* value corresponds with increased loading of the prosthetic limb. Perfectly balanced loading of both limbs is indicated at $DoA = 0$.

For the self-paced conditions, the data was segmented into separate sit-to-stand and stand-to-sit segments based on the derivative of $F_{z,feet}$, denoted as $\dot{F}_{z,feet}$. During sit-to-stand motions, $\dot{F}_{z,feet}$ starts at zero, increases to a peak and then returns to zero in an underdamped pattern. The start of a sit-to-stand motion was defined as the first instance where $\dot{F}_{z,feet}$ was greater than 300 N/s prior to the peak, and the end of the motion was determined when $|\dot{F}_{z,feet}|$ remained below 300 N/s for 60 ms after the peak. Likewise, the start of a stand-to-sit motion was defined as the next point where

$|\dot{F}_{z,feet}|$ returned to above 300 N/s, and the end of the stand-to-sit motion was defined where $|\dot{F}_{z,feet}|$ returned to below 300 N/s for 60 ms after the negative peak in $\dot{F}_{z,feet}$.

Using the segmented self-paced data, the average asymmetry during the motion $\overline{DoA}$ and RMS loading of the prosthetic leg was calculated and normalized by bodyweight $\bar{F}_{z,prosth}$. The average symmetry calculation was chosen over a symmetry calculation at the peak of the ground reaction force because such a peak does not always exist during stand-to-sit motions. The selected metrics quantify the user's ability to symmetrically load both legs during discrete motions, which should prevent detrimental compensations.

For the rapid sit-stand trials, the primary metrics were the time T_{5x} required to complete the task and average loading asymmetry $\overline{DoA}$ throughout the entire trial. No data segmenting was performed on that data. These metrics were selected to investigate whether the powered prosthesis allowed the user to complete the sit-stand motions faster and if symmetry was maintained during fast motions.

Linear mixed effects models were fit to each of the clinical metrics. The fixed effects included the categorical prosthesis type (powered or passive) and the continuous chair height, and the participant was treated as a random effect by including a subject-specific intercept. Statistically significant influences of the fixed effects were determined with appropriate Bonferroni corrections. As two comparisons were made for each independent set of data, significance was set to $p < 0.025$.

FIG. 5 illustrates the average phase (A), kinematic (B), and kinetic (C) trajectories observed across all self-paced trials (EX) with the tall stool height and the powered prosthesis. For comparison, FIG. 5 figure also illustrates the equivalent mean trajectories from able-bodied data (AB) collected at a chair height approximately equal to participant knee height, similar to the tall stool height. The shaded regions represent ± 1 standard deviation, where each sit-to-stand or stand-to-sit across movement all subjects was treated as an independent sample for both the able-bodied dataset and experimental results.

FIG. 5A illustrates a close correlation between the averaged able-bodied and experimental phase during the sit-to-stand motion, although the standard deviation in the phase variable is larger in the experimental data. During stand-to-sit, the experimental phase variable closely tracks the model dataset for the beginning of the motion, slightly overestimates during the last half of the motion.

FIG. 5B illustrates a high similarity between the experimental and average able-bodied joint angles, with the largest discrepancies occurring at the knee joint during the middle of the sit-to-stand motion, with some discrepancy in the ankle joint occurring at standing.

FIG. 5C illustrates a strong similarity between the experimental and able-bodied joint torques, with the largest discrepancy occurring at the end of the stand-to-sit motion, likely due to the overestimate in the phase variable.

Differences in $\overline{DoA}$ were observed between prosthesis types during the self-paced trials, as shown in FIG. 6A, where solid markers represent the powered prosthesis and hollow markers represent the participants' passive prostheses. During the stand-to-sit motion, the mixed effects model indicated a 0.361 reduction in $\overline{DoA}$ with the powered prosthesis, indicating that participants were significantly more symmetric compared to the passive prosthesis ($p \approx 10^{-56}$). Participant 2 showed the greatest symmetry improvement with the powered prosthesis during the stand-to-sit motion. The symmetry improvements with the powered prosthesis for sit-to-stand were less pronounced, with a reduction of 0.134, but this change was still significant ($p \approx 10^{-22}$). In both sit-to-stand and stand-to-sit motions, chair height had no significant effect on $\overline{DoA}$.

Similarly, significant improvements in $\bar{F}_{z,prosth}$ were observed with the powered prosthesis compared to the passive prostheses for both sit-to-stand ($p \approx 10^{-17}$) and stand-to-sit ($p \approx 10^{-52}$) in the self-paced trials. The RMS loading in FIG. 6B is $\bar{F}_{z,prosth}$ normalized by bodyweight. Similar to $\overline{DoA}$, this improvement was more pronounced during the stand-to-sit motions than during the sit-to-stand motions, with a 0.174 increase compared to a 0.061 increase in sit-to-stand. Again, the effect of chair height was insignificant. Interestingly, Participant 2 demonstrated little improvement with the powered prosthesis during the sit-to-stand motion while demonstrating the greatest improvement of all participants during the stand-to-sit motion. Conversely, Participants 1 and 3 demonstrated consistent improvement during both motions with the powered prosthesis.

With increased speed during the rapid trials, the participants still demonstrated reduced $\overline{DoA}$ by an average of 0.25 with the powered prosthesis, as illustrated in FIG. 7A ($p \approx 10^{-23}$). Here, the effect of chair height was insignificant. As illustrated in FIG. 7B, the time $\mathbb{T}_{5x}$ to complete one 5x sit-stand trial was also significantly improved with the powered prosthesis, with an average decrease of 3.29 seconds ($p \approx 10^{-8}$) and with Participant 2 showing the greatest improvement. The chair height

significantly affected completion time ($p \approx 10^{-5}$), with the shortest chair height requiring the most time and the tallest requiring the least time, on average.

In the sit-stand-walk trials, the controller smoothly switched between walk mode and sit-stand mode at the appropriate moments. The participants were able to intuitively change the prosthesis mode, allowing them to rise from one chair, walk across the room, and sit in the second chair without issue. Nonetheless, as illustrated in FIG. 8, the participants were, on average, faster with their passive prostheses than with the powered prosthesis.

In summary, a new control strategy has been developed for controlling relative movement of various parts of a powered prosthesis having at least one artificial joint. While most control strategies for powered prostheses are limited to rhythmic, periodic joint motion, the above-described control strategy uses impedance control to mimic human joint movement during transient tasks such as sitting from a standing position and/or standing from a seated position. Rather than divide up the transient motion and assigning constant values to the selected impedance parameters, the motion is treated as a continuous one such that the impedance parameters are continuously variable during the motion based on a phase variable, such as global thigh angle, that monotonically increases or decreases during the motion. The controller estimates the present phase in real-time based on that phase variable and on a model generated from able-bodied data. Each impedance parameter is varied according to the estimated phase and according to optimized models based on the able-bodied data as a function of phase. The need to “tune” the prosthesis for multiple discrete segments of the desired motion and/or for each individual user is thus eliminated. The controller can be seamlessly integrated with the earlier-described hybrid walking controller (HKIC), and possibly with other walking controllers.

Use of the thigh-based phase variable enables a single, continuous controller for sit-to-stand and stand-to-sit motions. The powered prosthesis may also include a high-level classifier that transitions between a walk mode and a sit-stand mode. The impedance model can incorporate dependency on knee velocity to decouple the sit-to-stand knee equilibrium angle trajectories from the stand-to-sit trajectories, reducing the model fitting errors by approximately half without the decoupling. The sit-stand control strategy demonstrates biomimetic joint kinematics and kinetics, as well as clinical benefits, including improved loading symmetry with the user’s natural leg and improved sit-stand motion speed relative to passive prostheses.

Comparisons between the powered prosthesis and able-bodied sit/stand data demonstrate that the controller generally produces biomimetic kinematics and kinetics (FIG. 5), with some noteworthy deviations. The largest discrepancies in ankle mechanics occur during standing ($s = 1$), though the magnitude of the discrepancy is quite small. The difference may be due to participant's electing to stand with their weight more towards the heel of the prosthesis, resulting in less dorsiflexion compared to the able-bodied dataset. Also, during the sit-to-stand motion, the prosthetic knee trajectories lead those from the able-bodied dataset in the middle of the motion, which is likely due to the experimental phase estimate leading the phase calculated from the able-bodied dataset. In contrast, during the stand-to-sit motion, the experimental phase lags that of the able-bodied dataset at the end of the motion, likely contributing to the excessive applied knee torque illustrated in FIG. 5(C). This excessive knee torque at the end of the stand-to-sit motion is the largest deviation from able-bodied mechanics, but it is unclear if this had a negative effect in the measured clinical outcomes. It may in fact improve clinical outcomes given that all participants were able to effectively use the biomimetic knee torque during stand-to-sit without any changes. Qualitative feedback from the participants, detailed further below, echoed this sentiment.

In comparison to participants' passive prostheses, the powered prosthesis significantly reduced average loading asymmetry, with increased prosthetic leg loading during sit-to-stand and stand-to-sit motions (FIGS. 6 and 7). Chair height did not have a significant effect on these results, suggesting that the controller is equally effective across multiple chair heights. In general, the improvements were more significant during stand-to-sit transitions compared to sit-to-stand transitions. One participant consistently loaded the powered prosthesis at least 50% more than their passive prosthesis and was able to achieve close to perfect symmetry during stand-to-sit with the powered prosthesis. In fact, the above-described sit-stand impedance controller reduced loading asymmetry by 79% to 0.09, on average, likely due at least in part to the larger magnitude of torque provided. The disclosed controller also increased RMS loading of the prosthesis to an average of 38% during sit-to-stand transitions and 46% during stand-to-sit transitions across all participants and chair heights. Given that loading asymmetry is known to correlate with osteoarthritis and lower back pain, this improved loading symmetry may reduce instances of those maladies in amputees. At a minimum, this new control strategy appears to reduce the user's overall reliance on their biological leg during sitting and standing movement, which has inherent benefits to the user.

It should be noted that the participants' prescribed device likely had a significant influence on both their baseline behavior and the effectiveness of the training protocol. For example, Participant 2's prescribed device was a Rheo Knee[®], which provides flexion resistance through the use of a magnetorheological damper. According to this participant, loading prevented the device from flexing the knee, forcing him to load most of his weight on the biological leg in order to sit down with his prescribed device.

Qualitative feedback from experiment participants was also obtained. In general, all participants liked using the powered prosthesis and found it helpful compared to their passive devices. One noted that "standing up from the short chair was way harder" with his passive device compared to the powered prosthesis. Another noted that "it's nice that I don't have to predict where my rear is going to go" during stand-to-sit with the powered prosthesis, because his passive device "just releases, it's just a guess." Although fatigue was not directly measured or quantified, one participant noted that "tired-wise, I think [the powered prosthesis] is helpful... but overall I'm fatiguing during the whole experiment."

The disclosed controller is not limited to the specific impedance parameters or optimization techniques described above. For example, the impedance parameters may be optimized over multiple chair heights with additional able-bodied data. Or the impedance parameters may be parameterized as functions of chair height. Other variables may be identified that affect the biomimetics of the prostheses, and the impedance parameters may be optimized over multiple values of those variables and/or parameterized as functions of those variables.

It is to be understood that the foregoing is a description of one or more embodiments of the invention. The invention is not limited to the particular embodiment(s) disclosed herein, but rather is defined solely by the claims below. Furthermore, the statements contained in the foregoing description relate to particular embodiments and are not to be construed as limitations on the scope of the invention or on the definition of terms used in the claims, except where a term or phrase is expressly defined above. Various other embodiments and various changes and modifications to the disclosed embodiment(s) will become apparent to those skilled in the art. All such other embodiments, changes, and modifications are intended to come within the scope of the appended claims.

As used in this specification and claims, the terms "e.g.," "for example," "for instance," "such as," and "like," and the verbs "comprising," "having," "including," and their other verb forms, when used in conjunction with a listing of one or more components or other items, are each to be construed as open-ended, meaning that the listing is not to be considered as excluding other, additional
5 components or items. Other terms are to be construed using their broadest reasonable meaning unless they are used in a context that requires a different interpretation.

CLAIMS

1. A powered prosthesis comprising a joint and an impedance controller configured to continuously vary impedance at the joint during a user transition between a standing position and a seated position.
2. The powered prosthesis of claim 1, wherein the impedance at the joint is a continuous function of one or more impedance parameters, each impedance parameter being a continuous function of a sit-stand phase defined between the standing position and the seated position.
3. The powered prosthesis of claim 2, wherein each impedance parameter function is optimized based at least in part on able-bodied data independent from the user.
4. The powered prosthesis of claim 2, wherein the controller estimates the sit-stand phase in real-time.
5. The powered prosthesis of claim 4, wherein the controller estimates the sit-stand phase based at least in part on a phase variable that monotonically increases or decreases during the user transition.
6. The powered prosthesis of claim 5, wherein the phase variable is a thigh angle of the user.
7. The powered prosthesis of claim 3, wherein at least one of the one or more impedance parameters is also a function of a decoupling variable which is zero when the user transition is from the standing position to the seated position and non-zero when the user transition is from the seated position to the standing position.
8. The powered prosthesis of claim 7, wherein an equilibrium angle of the joint is one of the impedance parameters and the decoupling variable is an angular velocity of the joint.
9. The powered prosthesis of claim 8, wherein the joint is a knee joint.
10. The powered prosthesis of claim 1, wherein the impedance controller is configured to

continuously vary impedance at the joint both during a user transition from the standing position to the seated position and during a user transition from the seated position to the standing position.

11. A powered knee-ankle prosthesis according to claim 1, wherein the joint is a knee joint and the prosthesis further comprises an ankle joint, the controller being configured to continuously vary impedance at both joints during the user transition.

12. The knee-ankle prosthesis of claim 11, further comprising a hybrid walking controller configured to control impedance at both joints during a stance phase of walking and to control kinematics of both joints during a gate phase of walking, wherein both controllers control impedance as a function of a thigh angle of the user.

13. The knee-ankle prosthesis of claim 12, wherein the prosthesis is configured to transition between a sit-stand mode and a walk mode based at least in part on a thigh angle and a thigh angular velocity of the user.

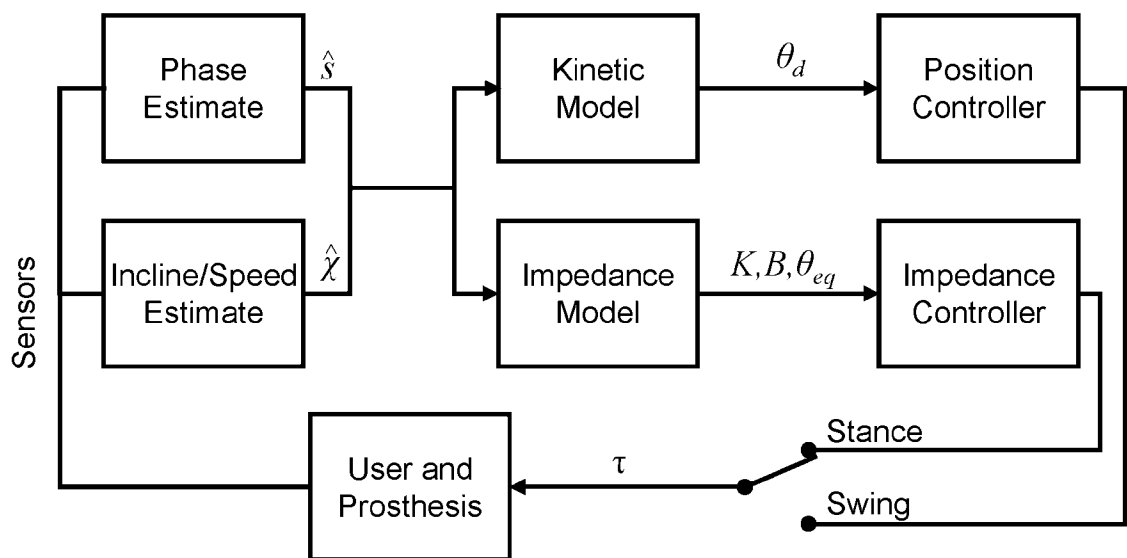


FIG. 1

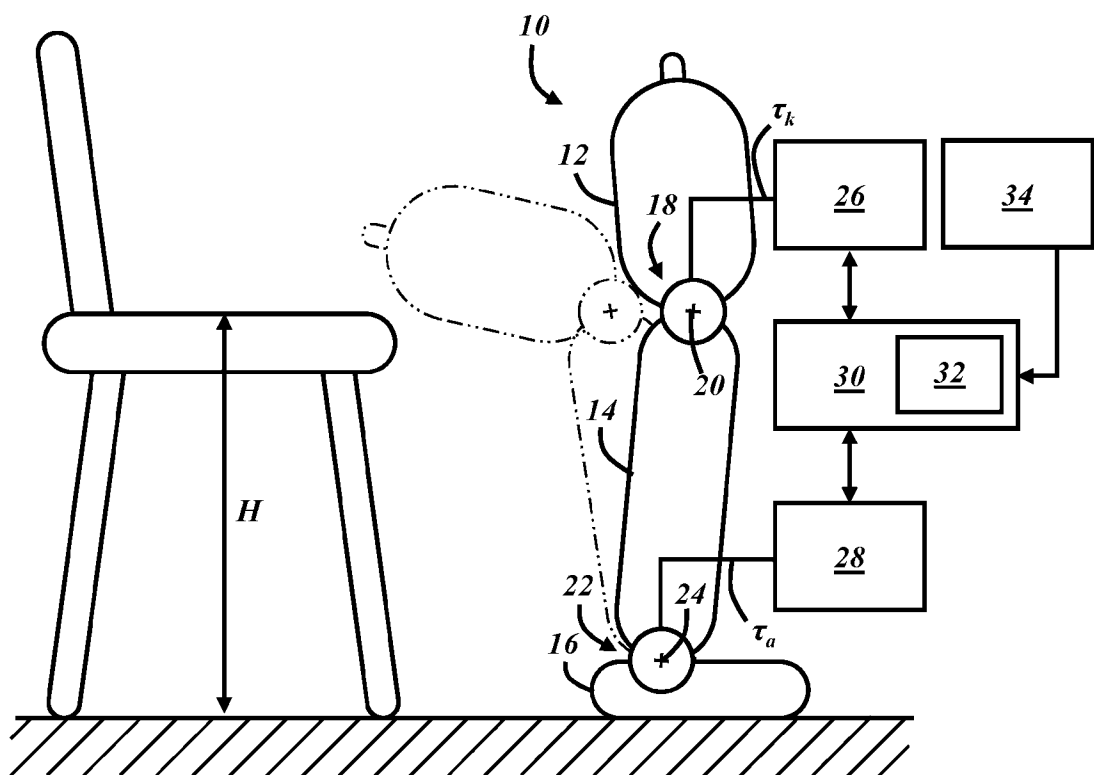


FIG. 2

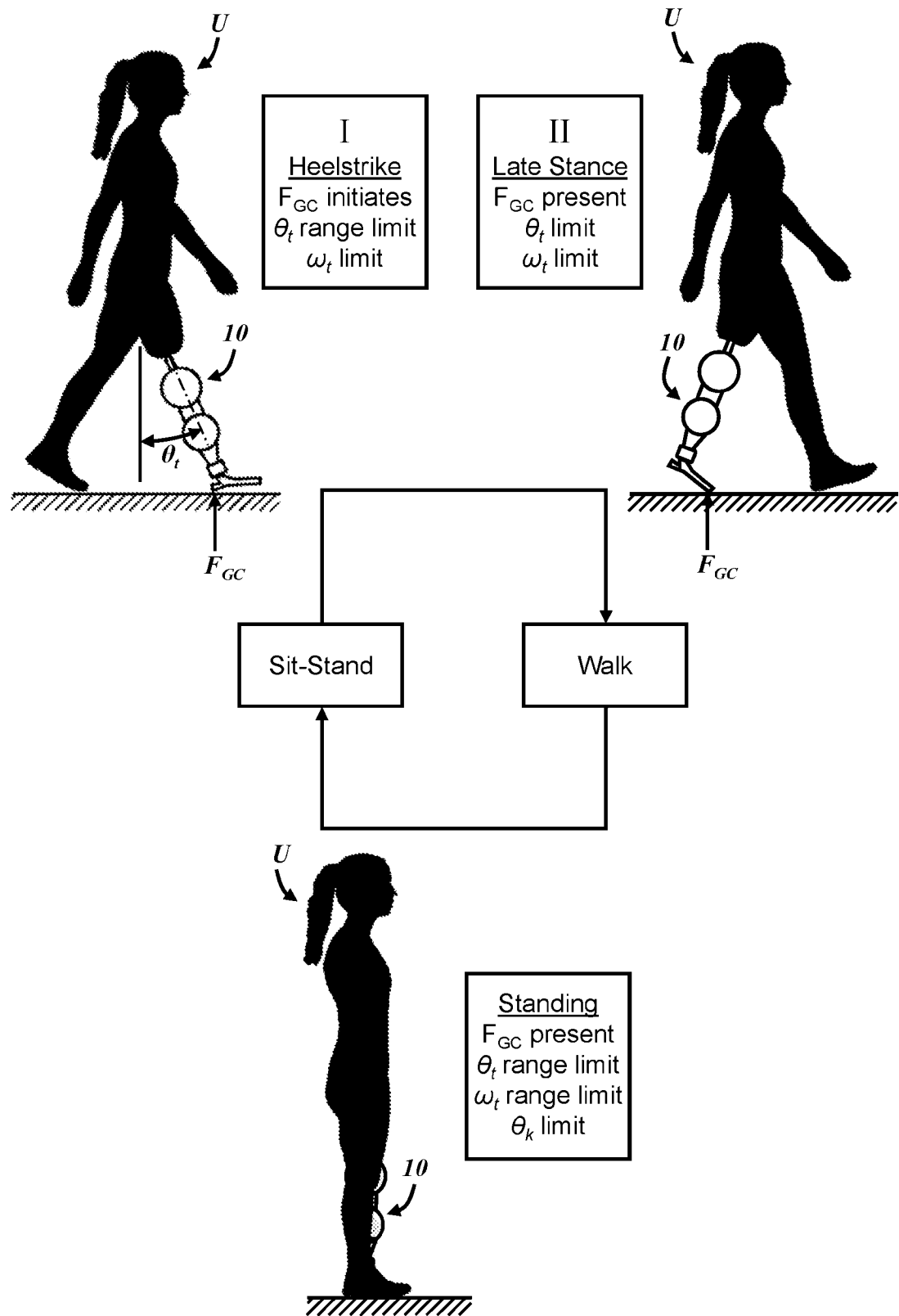


FIG. 3

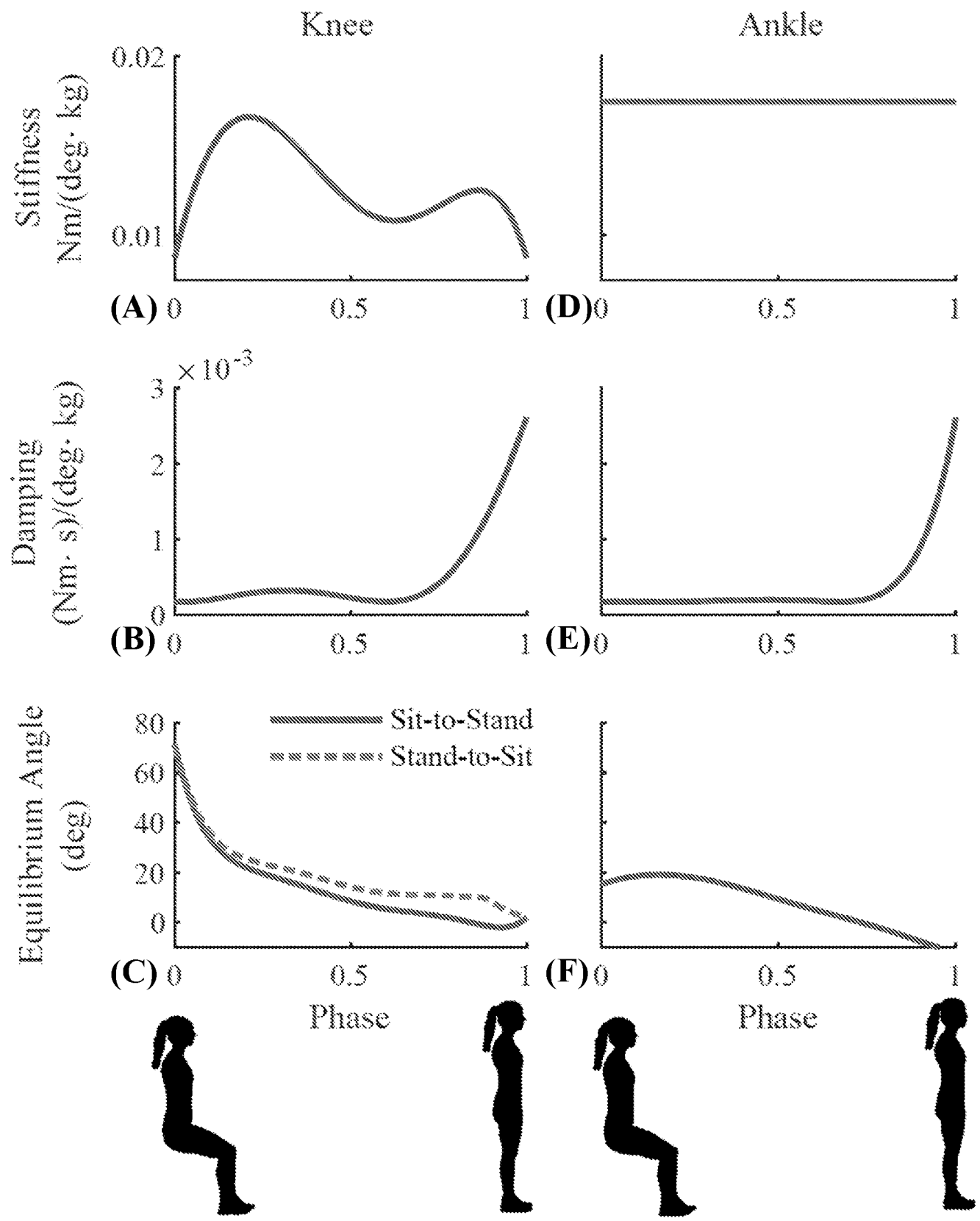
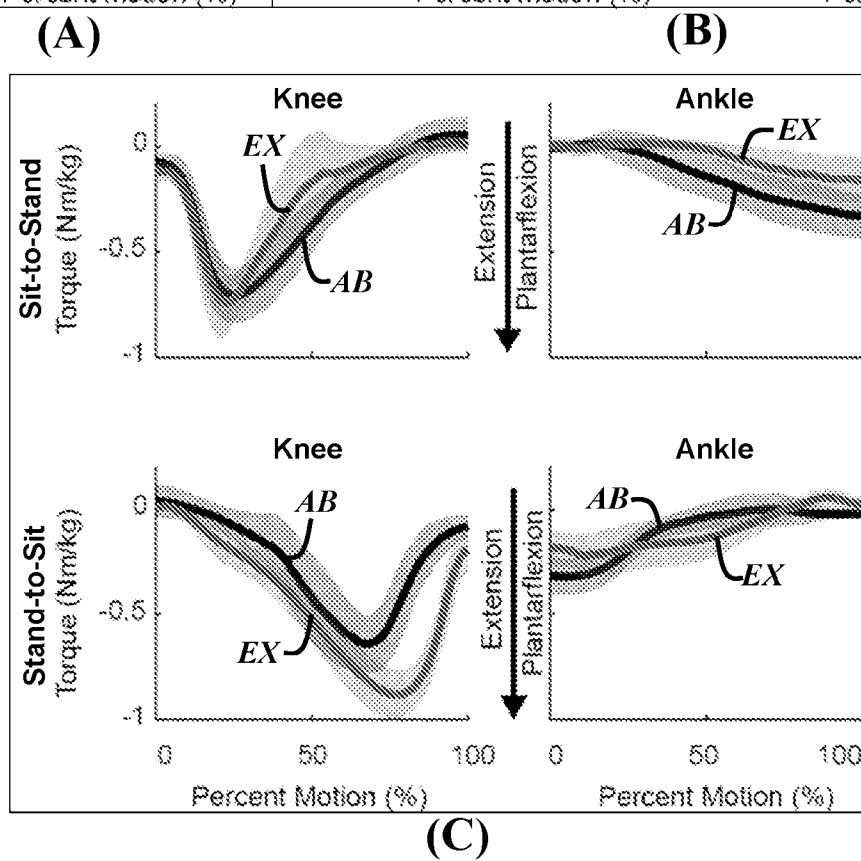
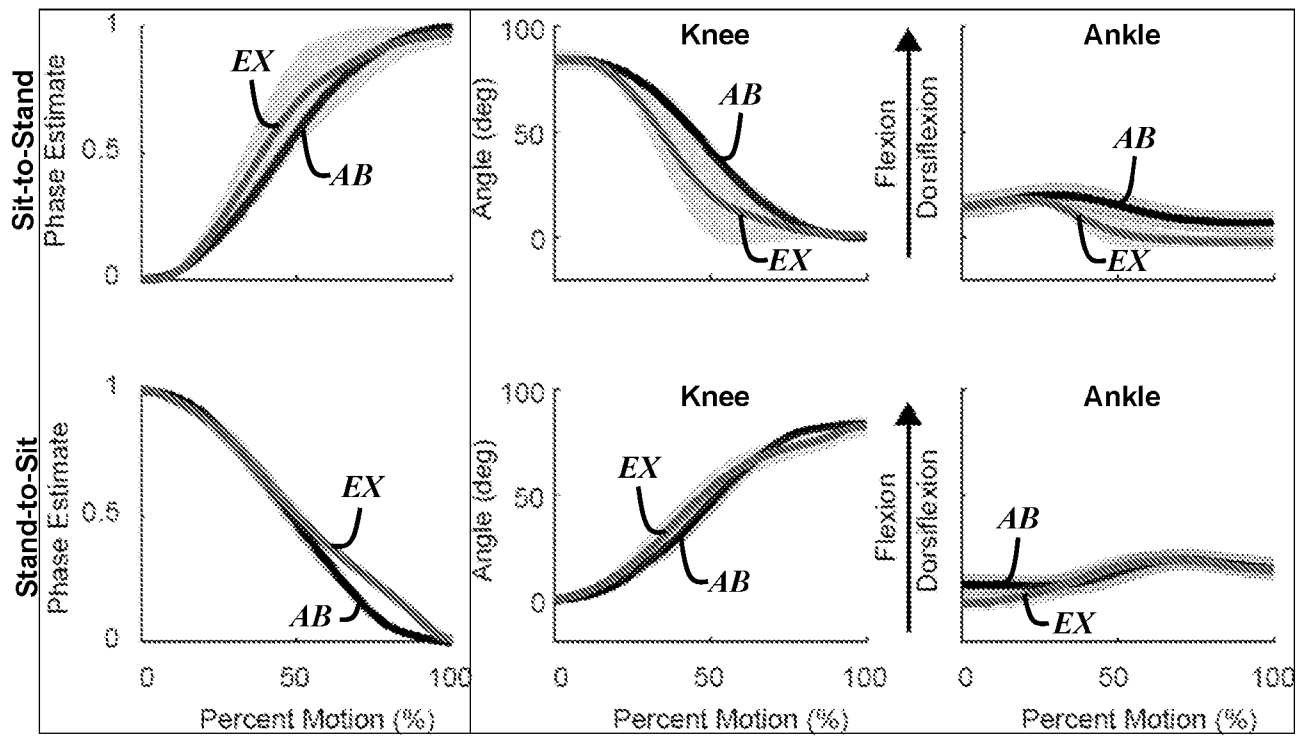


FIG. 4

**FIG. 5**

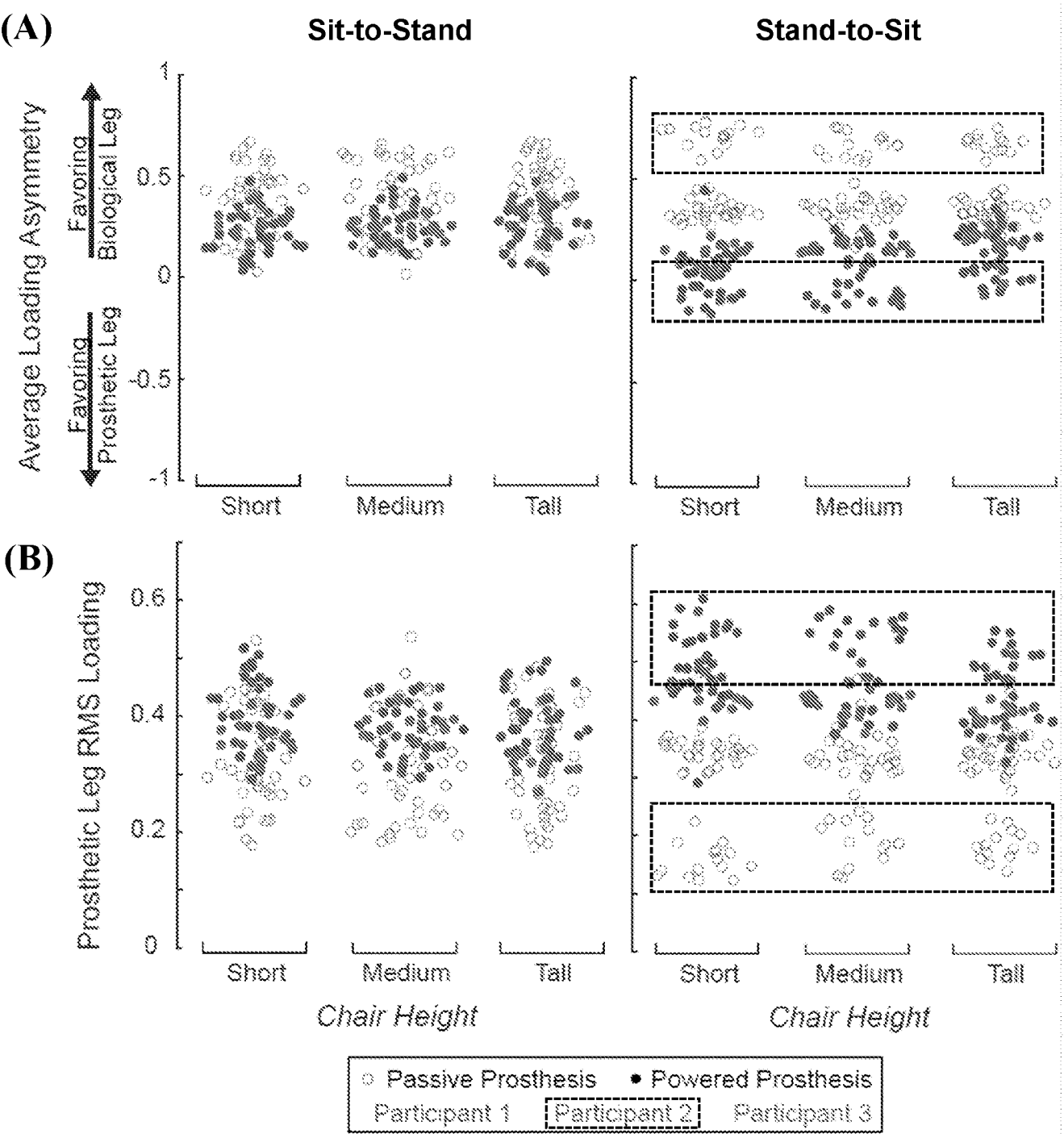
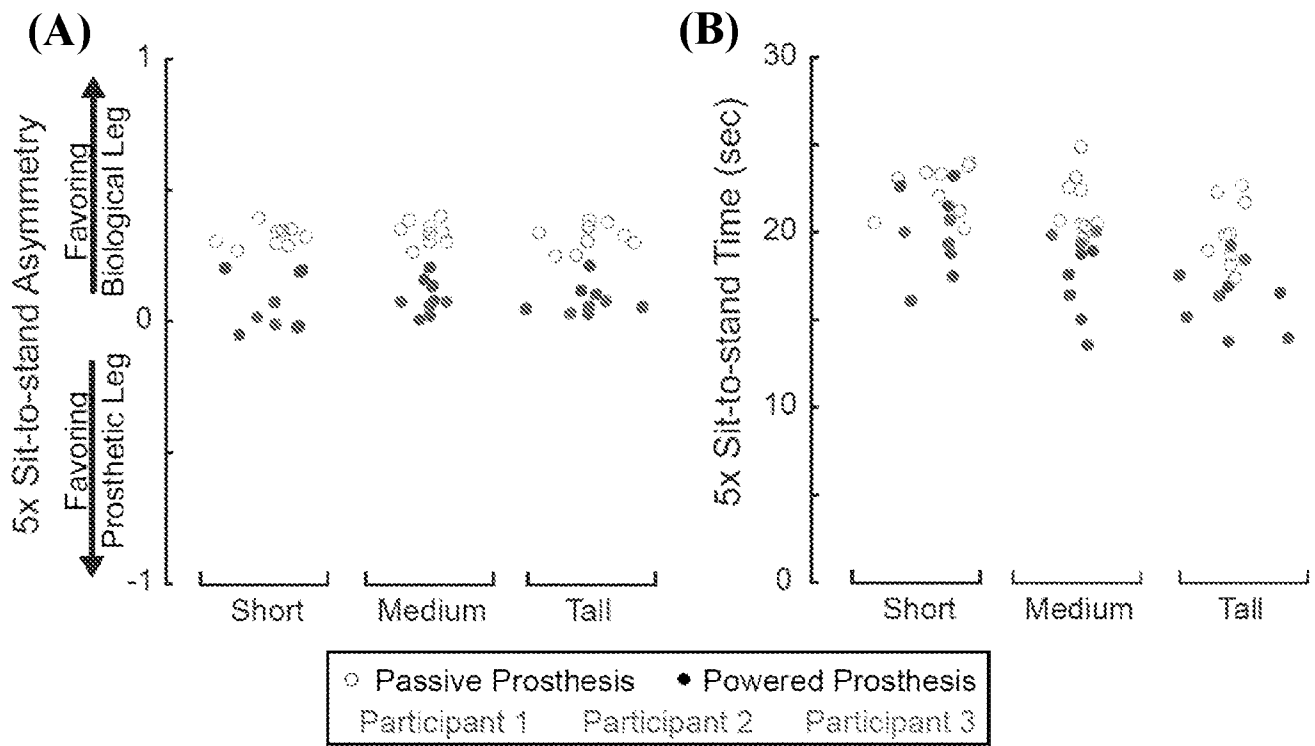
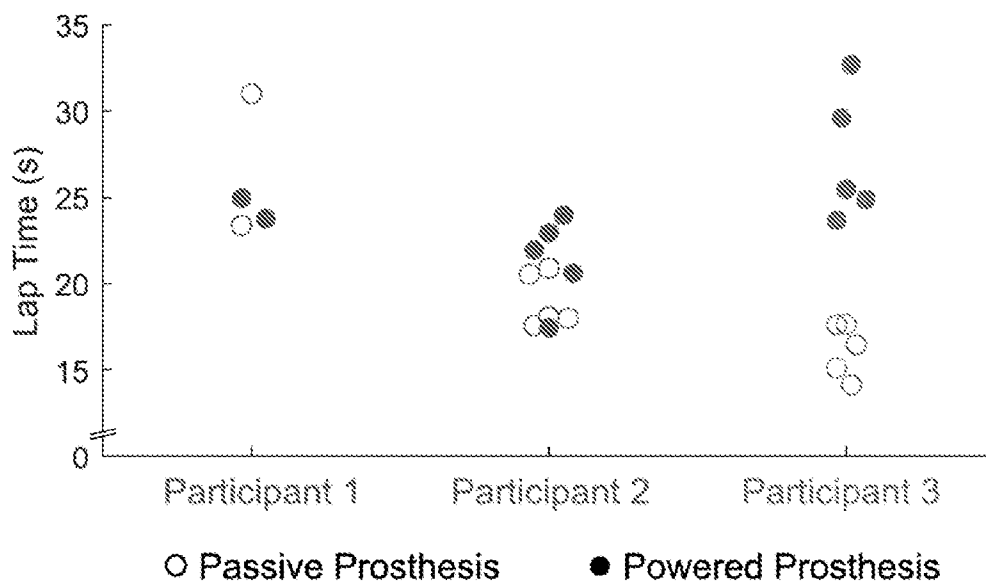


FIG. 6

**FIG. 7****FIG. 8**

INTERNATIONAL SEARCH REPORT

International application No.

PCT/US2024/035847

A. CLASSIFICATION OF SUBJECT MATTER

A61F 2/76(2006.01)i; A61F 2/50(2006.01)i; A61F 2/60(2006.01)i; A61F 2/68(2006.01)i

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

A61F 2/76(2006.01); A61F 2/60(2006.01); A61F 2/64(2006.01); A61F 2/66(2006.01); A61F 2/68(2006.01);
A61F 2/70(2006.01); A61G 5/14(2006.01); B25J 11/00(2006.01)

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Korean utility models and applications for utility models
Japanese utility models and applications for utility models

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

eKOMPASS(KIPO internal) & Keywords: prosthesis, joint, impedance, angle, seated, stand, walk, continuous function

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	WELKER, C. G. et al., "Data-driven variable impedance control of a powered knee-ankle prosthesis for sit, stand, and walk with minimal tuning", IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS), 2022, pages 9660-9667(Author Manuscript version pages 1-19) pages 3-5, 12	1-13
A	US 2015-0127118 A1 (IWALK, INC.) 07 May 2015 (2015-05-07) claim 1; paragraphs [0047], [0071]	1-13
A	WO 2022-087161 A1 (UNIVERSITY OF UTAH RESEARCH FOUNDATION) 28 April 2022 (2022-04-28) claims 1, 2	1-13
A	KR 10-2012-0107928 A (IWALK, INC.) 04 October 2012 (2012-10-04) claims 1-10	1-13
A	JP 2018-038785 A (PANASONIC IP MANAGEMENT CORP.) 15 March 2018 (2018-03-15) claims 1-19	1-13



Further documents are listed in the continuation of Box C.



See patent family annex.

* Special categories of cited documents:

"A" document defining the general state of the art which is not considered to be of particular relevance

"D" document cited by the applicant in the international application

"E" earlier application or patent but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search

22 October 2024

Date of mailing of the international search report

23 October 2024

Name and mailing address of the ISA/KR

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INTERNATIONAL SEARCH REPORT
Information on patent family members

International application No.

PCT/US2024/035847

Patent document cited in search report			Publication date (day/month/year)	Patent family member(s)			Publication date (day/month/year)
US	2015-0127118	A1	07 May 2015	CA	2876187	A1	19 December 2013
				CA	2876187	C	26 January 2021
				EP	2858602	A2	15 April 2015
				US	10531965	B2	14 January 2020
				US	11707364	B2	25 July 2023
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