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The Effects of Prosthesis Inertial Properties on Prosthetic Knee Moment and Hip Energetics Required to Achieve Able-bodied Kinematics

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***Abstract*—There is a major need in the developing world for a low-cost prosthetic knee that enables users to walk with able-bodied kinematics and low energy expenditure. To efficiently design such a knee, the relationship between the inertial properties of a prosthetic leg and joint kinetics and energetics must be determined. In this paper, using inverse dynamics, the theoretical effects of varying the inertial properties of an above-knee prosthesis on the prosthetic knee moment, hip power, and absolute hip work required for walking with able-bodied kinematics were quantified. The effects of independently varying mass and moment of inertia of the prosthesis, as well as independently varying the masses of each prosthesis segment, were also compared. Decreasing prosthesis mass to 25% of physiological leg mass increased peak late-stance knee moment by 43% and decreased peak swing knee moment by 76%. In addition, it reduced peak stance hip power by 26%, average swing hip power by 76%, and absolute hip work by 22%. Decreasing upper leg mass to 25% of its physiological value reduced absolute hip work by just 2%, whereas decreasing lower leg and foot mass reduced work by up to 22%, with foot mass having the greater effect. Results are reported in the form of parametric illustrations that can be utilized by researchers, designers, and prosthetists. The methods and outcomes presented have the potential to improve prosthetic knee component selection, facilitate able-bodied kinematics, and reduce energy expenditure for users of low-cost, passive knees in developing countries, as well as for users of advanced active knees in developed countries.**

***Index Terms*—prosthesis mass, prosthesis moment of inertia, inverse dynamics, prosthetic knee moment, hip power, hip work, design for the developing world, India.**

I. Introduction

In recent decades, the evolution of passive mechanical prosthetic knees into active electromechanical devices has enabled above-knee amputees to walk with improved kinematics and energy expenditure. Unfortunately, microprocessor-controlled active prosthetic knees are prohibitively expensive for the majority of amputees living in the developing world. We aim to design prosthetic knees that allow users to walk with able-bodied kinematics and reduced energy expenditure, but are affordable in developing countries. In this work, we determine the prosthetic knee moments and hip energetics required for amputees to walk with able-bodied kinematics, as well as the sensitivity of these parameters to the inertial properties of the prosthetic leg.

According to the International Society for Prosthetics and Orthotics and the World Health Organization, approximately 30 million people worldwide are in need of prosthetic and orthotic devices [1], [2], [3]. Our current project focuses on developing a prosthetic knee for use in India, with the intent of disseminating it to other developing countries in the future. Currently, there are approximately 230,000 above-knee amputees living in India [4], [5]. Since many of these individuals experience poverty [6], lose their jobs, and face social discrimination [3] because of their disability, they have a major need for a low-cost prosthetic knee that allows them to walk normally, perform work, and appear able-bodied. Unfortunately, commonly available prosthetic knees in developing countries [7], [8] have been found to have one or more of the following inadequacies: inhibition of normative gait, mechanical failures, and low user-satisfaction [9], [10], [8], [11]. Most prosthetic knees available in the United States and Europe are too expensive for amputees in developing countries, with microprocessor controlled knees and powered knees costing over $50,000 [12].

From a technical perspective, beyond facilitating safe and stable locomotion, two important goals of above-knee prosthesis design are 1) to enable above-knee amputees to walk with able-bodied kinematics, and 2) to minimize their metabolic energy expenditure. Above-knee amputees using commercially available passive and active knees often do not walk with able-bodied kinematics [13], [14]. Furthermore, unilateral transfemoral amputees consume 20%-119% more oxygen per unit distance than able-bodied controls [15], [16], and bilateral transfemoral amputees consume 52%-280% more oxygen per unit distance than able-bodied controls [17], [18]. To achieve able-bodied kinematics, researchers have designed prosthetic knees to produce moments that facilitate normative motions [19], [20]. In an attempt to lower metabolic energy expenditure, researchers have typically reduced inertial properties (e.g., mass and moment of inertia) of the prosthetic leg [21].

Each of these design approaches has distinct limitations. First, researchers aiming to achieve able-bodied kinematics have typically optimized prosthetic knee components to produce the knee moments generated by able-bodied humans walking with normative kinematics [19], [20]. However, since a physiological leg typically weighs more than a prosthetic leg [22], the prosthetic knee moment required for an above-knee prosthesis user to walk with able-bodied kinematics may be significantly different. Second, the effects of prosthesis inertial properties on energy expenditure have not yet been completely determined. Researchers aiming to experimentally or theoretically quantify the effects of prosthesis inertial alterations on energy expenditure have typically applied mass perturbations (i.e., physical or simulated masses) to the prosthesis and determined metabolic or mechanical energy expenditure [22],

[23], [24], [25], [26], [27]. Mass perturbations alter both mass and mass distribution (which in turn affects moment of inertia), confounding the effects of these parameters. Furthermore, few studies have compared the effects of adding mass to particular segments of an above-knee prosthesis (i.e., socket, shank, and foot). Thus, it is difficult to predict how changing the inertial properties of a particular segment independently of the others would affect energy expenditure.

The preceding design limitations may not be highly critical for advanced active knees, which often contain electromechanical components (e.g., batteries, microprocessors, actuators, and sensors) and control algorithms that can compensate for undesired kinematics. For instance, users of active knees designed by Sup et al. [20] and Martinez-Villalpando and Herr [19] exhibited satisfactory kinematics even though the components and/or control schemes were optimized to reproduce able-bodied knee moments (rather than the knee moments required for prosthesis user to walk with able-bodied kinematics). However, the design limitations are critical for prosthetic knees for developing countries, which cannot use electromechanical components due to the expenses of maintenance, replacement, and charging batteries.

In this paper, we address these limitations. We use inverse dynamics to theoretically calculate the effects of independently varying prosthesis mass and moment of inertia on the knee moment, hip power, and absolute hip work (a measure of mechanical energy expenditure at the hip, which is a primary actuator of an above-knee prosthesis) required for walking with able-bodied kinematics. In addition, we calculate the effects of independently varying the masses of particular segments on absolute hip work. These kinetic and energetic parameters are computed over the entire gait cycle. The results are analyzed in the context of prosthetic knee design and are reported in the form of parametric illustrations that can be readily used by designers, prosthetists, and researchers. Our methods and outcomes have potential to improve prosthetic knee component selection and optimization for both passive and active knees, facilitate able-bodied kinematics, and reduce energy expenditure of above-knee amputees living in developing and developed nations. We recognize that achieving able-bodied kinematics and reduced energetics, which is the focus of this paper, are not the only goals of a prosthesis design. Ideally, additional requirements would have to be considered such as stability, safety, aesthetics, maintenance and ease of repair.

II. Methods

To determine the effects of prosthesis inertial properties on knee moment and hip energetics, the following steps were taken: 1) a model of a prosthetic leg was designed, 2) dimensions were prescribed, 3) inertial properties were varied, 4) kinematics were prescribed, 5) external forces were calculated, 6) inverse dynamics was used to calculate knee moment, and 7) hip power and absolute hip work were computed. Each of these steps is described in detail below. All calculations were conducted in MATLAB (R2012a, The MathWorks, Natick, MA [28]). Due to the scarcity of publicly available complete

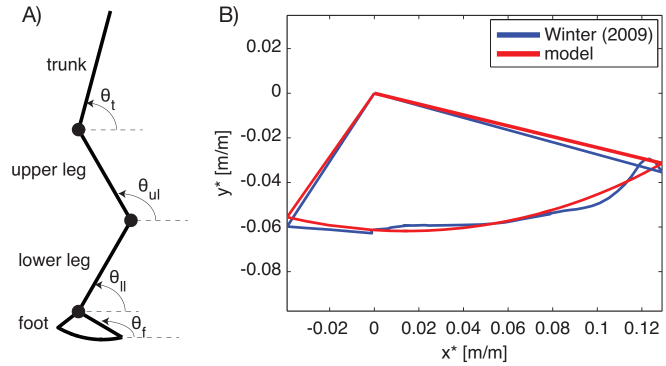

Fig. 1. 2-dimensional, 4-segment model of prosthetic leg. A) Model of prosthetic leg, where θ_t is trunk angle, θ_{ul} is upper leg angle, θ_{ll} is lower leg angle, and θ_f is foot angle. B) Roll-over model of foot. Foot roll-over shape computed from [29] is compared to circular approximation used in model. Ankle joint is located at origin. Symbol x^* designates x-coordinate normalized to body height, and symbol y^* designates y-coordinate normalized to body height. $R^2 = 0.93$.

gait data-sets in literature for a large population of able-bodied adults, kinematic and kinetic data from multiple sources were estimated and combined (as discussed in each of the following sections).

A. Design of the model

A 2-dimensional, 4-segment link-segment model was designed to model the prosthetic leg of a unilateral transfemoral amputee wearing an above-knee prosthesis (Fig. 1A). The model consisted of a trunk segment, an upper leg segment (residual limb and socket), a lower leg segment (shank), and a foot segment. To model the foot segment, sample center of pressure (COP) data were acquired from [29] for walking at a fast cadence (approximately 125 steps/min [30]). The COP data were transformed into the reference frame of the foot to compute a foot roll-over shape [31], and a circular arc was fitted to the data (Fig. 1B).

The COP data were not reported for natural cadence in [30] and were acquired for fast cadence from [29]. The roll-over shape has been shown to be independent of cadence speed [31]. Hence, the foot roll-over shape computed here with COP data for fast cadence was implemented for able-bodied kinematics at natural cadence (as discussed in section 2.4).

B. Dimensions of the model

Lengths and centers of mass (COM) of the segments were prescribed according to anthropometric ratios of able-bodied humans [32], [29] scaled to the average body height of subjects (1.75 m) in the joint-angle data set described later [30]. These values were held constant and are listed in Table I.

C. Inertial properties of the model

To determine the effects of independently varying prosthesis mass and moment of inertia on knee moment and hip energetics, the following inertial alterations were applied to the model:

TABLE I
DIMENSIONS AND INERTIAL PROPERTIES OF THE MODEL. CENTERS OF MASS ARE EXPRESSED AS DISTANCE FROM CORRESPONDING PROXIMAL JOINT. MOMENTS OF INERTIA ARE EXPRESSED WITH RESPECT TO GIVEN CENTERS OF MASS

	Length $[m]$	Center of mass $[m]$	Mass $[kg]$	Moment of inertia $[kg * m^2]$
Upper leg	0.4288	0.1856	6.910	0.1325
Lower leg	0.4305	0.1864	3.213	0.0543
Foot	0.1369	0.0684	1.002	0.0042

1) Decreasing the masses and moments of inertia of all segments by the same factor
2) Decreasing the moments of inertia of all segments by the same factor, but holding the masses constant
3) Decreasing the masses of all segments by the same factor, but holding the moments of inertia constant

The preceding inertial alterations have the following physical analogues, respectively:

1) Decreasing the material density of all segments
2) Moving the mass of each segment closer to its COM
3) Decreasing mass at the COM of each segment

Masses and moments of inertia of the segments were decreased to 25%, 50%, 75%, and 100% of their corresponding able-bodied values. This range was chosen to test configurations with higher and lower inertial properties than what a typical above-knee prosthesis user would have, with approximately 50% upper leg mass and 33% lower leg and foot mass compared to able-boded values [22], [13]. Able-bodied masses and moments of inertia were determined by scaling anthropometric ratios of able-bodied humans [33], [34], [29] to the average body mass of subjects (69.1 kg) in the joint-angle data set described in the next section [30]. These values are listed in Table I.

In addition, to determine the effects of independently altering the masses of particular prosthesis segments on absolute hip work, the mass of each segment was varied between 25% and 100% of its corresponding able-bodied value, with 4 data points distributed evenly throughout the range for upper leg mass and 25 data points distributed throughout the range for lower leg mass and foot mass. All possible combinations of upper leg mass, lower leg mass, and foot mass were evaluated.

D. Kinematics of the model

Since the study aimed to calculate the prosthetic knee moment and hip energetics required for walking with able-bodied kinematics at a natural cadence, corresponding kinematics were applied to the model. Because a complete kinematic data set for a large population of able-bodied humans could not be found in the literature, kinematic data were extrapolated and combined from multiple sources. Able-bodied joint angles were acquired from [30], which reported average joint angles for 19 able-bodied adults walking at a natural cadence (105 steps/min). Hip position was approximated to be stationary due to small velocities during gait [35]. Trunk angle was acquired from [29] for walking at a fast cadence. Thorstensson [36] determined that the range of angular displacement of the trunk does not change with walking speed, but that the timing changes by -10% of the gait cycle from slow to fast walking. In order to represent walking at a natural cadence, the trunk angle data were shifted in time by approximately +5% of the gait cycle to represent walking at a natural cadence. Finally, COP was estimated from the foot roll-over shape. The data used to generate the roll-over shape [30] was recorded for walking at a fast cadence. Hansen et al. [31] determined that a circular arc fitted to a roll-over shape based on the knee, ankle, and foot does not change significantly with walking speed, and it was assumed that the same result applies to roll-over shape based on the foot alone. The COP at any time in stance was determined by identifying the lowest point of the foot segment at that time.

E. External forces on the model

Only two external forces act on the body during normal walking: gravity and the ground reaction force (GRF). From multi-rigid-body dynamics, the net external force on a system is equal to the sum of the mass-acceleration products of all the bodies. Applying these observations to a link-segment model of the human body (Fig. 1A), the net GRF can be calculated as

$$\overrightarrow{GRF} = \sum_{i=1}^{N} m_i(\ddot{\overrightarrow{r}}_{i\,COM} + g\hat{y}) \tag{1}$$

where N is the number of segments representing the body, m_i is the mass of the i^{th} segment, $\ddot{\overrightarrow{r}}_{i\,COM}$ is the position vector of the COM of the i^{th} segment relative to the origin of an inertial reference frame, g is the gravitational constant, and $\hat{y}$ is a unit vector in the positive vertical direction.

The total GRF acting on a unilateral transfemoral amputee ($\overrightarrow{GRF}_{amp}$) using an above-knee prosthesis to walk with able-bodied kinematics is equal to the total GRF acting on an able-bodied human ($\overrightarrow{GRF}_{able}$), minus the fraction of able-bodied GRF that acts on the additional leg mass of a physiological leg compared to a prosthetic leg. Thus, $\overrightarrow{GRF}_{amp}$ can be computed as

$$\begin{aligned}\overrightarrow{GRF}_{amp} = \Bigg[& \overrightarrow{GRF}_{able} \\ & - \begin{pmatrix} (m_{ul} - m_{ul_{pr}})(\ddot{\overrightarrow{r}}_{ul_{COM}} + g\hat{y}) + \\ (m_{ll} - m_{ll_{pr}})(\ddot{\overrightarrow{r}}_{ll_{COM}} + g\hat{y}) + \\ (m_f - m_{f_{pr}})(\ddot{\overrightarrow{r}}_{f_{COM}} + g\hat{y}) \end{pmatrix} \Bigg]\end{aligned} \tag{2}$$

where m_{ul}, m_{ll}, and m_f are the masses of the upper leg, lower leg, and foot in an able-bodied human, $m_{ul_{pr}}$, $m_{ll_{pr}}$, $m_{f_{pr}}$ are the masses of the same segments in a prosthetic leg, and $\overrightarrow{r}_{ul_{COM}}$, $\overrightarrow{r}_{ll_{COM}}$, and $\overrightarrow{r}_{f_{COM}}$ are the position vectors of the COMs of the upper leg, lower leg, and foot relative to the origin of an inertial reference frame. $\overrightarrow{GRF}_{able}$ was acquired from [30]. It was assumed that the COMs of the able-bodied segments were preserved in the prosthetic leg segments.

During single support, the GRF acting on the prosthetic leg ($\overrightarrow{GRF}_{pr}$) is equal to the total GRF acting on the amputee

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($\overrightarrow{GRF}_{amp}$). During the double-support phases, $\overrightarrow{GRF}_{amp}$ is distributed between the two legs, and $\overrightarrow{GRF}_{pr}$ is indeterminate. During early-stance double support, $\overrightarrow{GRF}_{pr}$ was approximated by a linear interpolation from zero to the initial single-support value of $\overrightarrow{GRF}_{amp}$. During late-stance double support, $\overrightarrow{GRF}_{pr}$ was approximated by a linear interpolation from the final single-support value of $\overrightarrow{GRF}_{amp}$ to zero. The length of double-support was estimated to be 12.5% of the gait cycle [37], [35].

F. Calculation of joint moments

A standard 2-dimensional inverse dynamics procedure [38] was used to compute joint moments, including knee moment. Because able-bodied kinematics were applied to the model, this knee moment is the prosthetic knee moment required to walk with able-bodied kinematics. The moment values could alternatively have been found by conducting a forward dynamics simulation of the model and varying joint moments until achieving able-bodied kinematics. However, this iterative approach is unnecessary when the direct approach of inverse dynamics is feasible. In addition, musculoskeletal simulation software such as OpenSim [39] could have been used to perform inverse dynamics. A manual computation of inverse dynamics was preferred in order to have simple and direct access to all intermediate values.

To evaluate the kinematic and kinetic approximations used in our model, prosthesis inertial properties were initially set to corresponding able-bodied values, and resulting joint moments were compared to those reported in [30] (Fig. 2). The model closely matched reported values, with moderate accuracy for the hip and high accuracy for the knee and ankle ($R^2 > 0.90$).

G. Calculation of hip energetics

Since the hip is the primary physiological actuator of a prosthetic leg, hip energetics were examined in this study. Hip power (P_{hip}) was calculated by multiplying hip moment by hip angular velocity. In order to compute a measure of mechanical energy expenditure with physiological relevance, absolute joint work was calculated. Absolute joint work is defined as

$$W_{joint}^{abs} = \int_{t_0}^{t_{100}} |P_{joint}| dt \quad (3)$$

where W_{joint}^{abs} is the absolute work at a given joint, t_0 is the time at 0% gait cycle (heel strike), t_{100} is the time at 100% gait cycle (heel strike of the ipsilateral foot), and P_{joint} is the power at the joint. Muscles expend metabolic energy during both concentric and eccentric contractions, and the value of absolute joint work increases during both concentric and eccentric motions of the joint [40].

III. Results

A. Effects of prosthesis mass and moment of inertia on knee moment

The effects of prosthesis inertial variations on knee moment are illustrated in Fig. 3. Decreasing both masses and moments

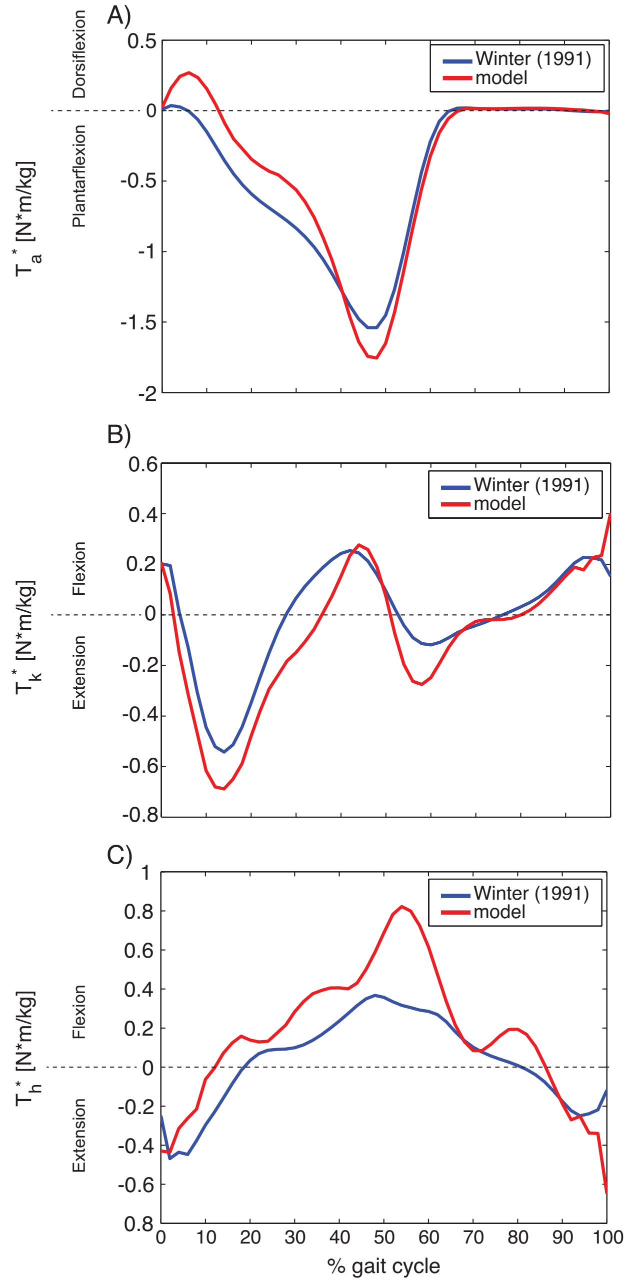

Fig. 2. Comparison between moments calculated by the model and moments reported in [30]. A) Symbol T_a^* designates ankle moment normalized to body mass. $R^2 = 0.93$. B) Symbol T_k^* designates knee moment normalized to body mass. $R^2 = 0.91$. C) Symbol T_h^* designates hip moment normalized to body mass. $R^2 = 0.80$.

of inertia of all segments of the prosthetic leg had a large effect on the knee moment required for able-bodied kinematics, increasing the peak magnitude of knee extension moment during late stance by up to 43% and decreasing the peak magnitude of swing flexion moment by up to 76%. Decreasing masses and holding moments of inertia constant had a similar effect, increasing the peak magnitude of late stance extension moment by up to 43% and decreasing the peak magnitude of swing flexion moment by up to 60%. Finally, decreasing moments of inertia and holding masses constant had a negligible effect

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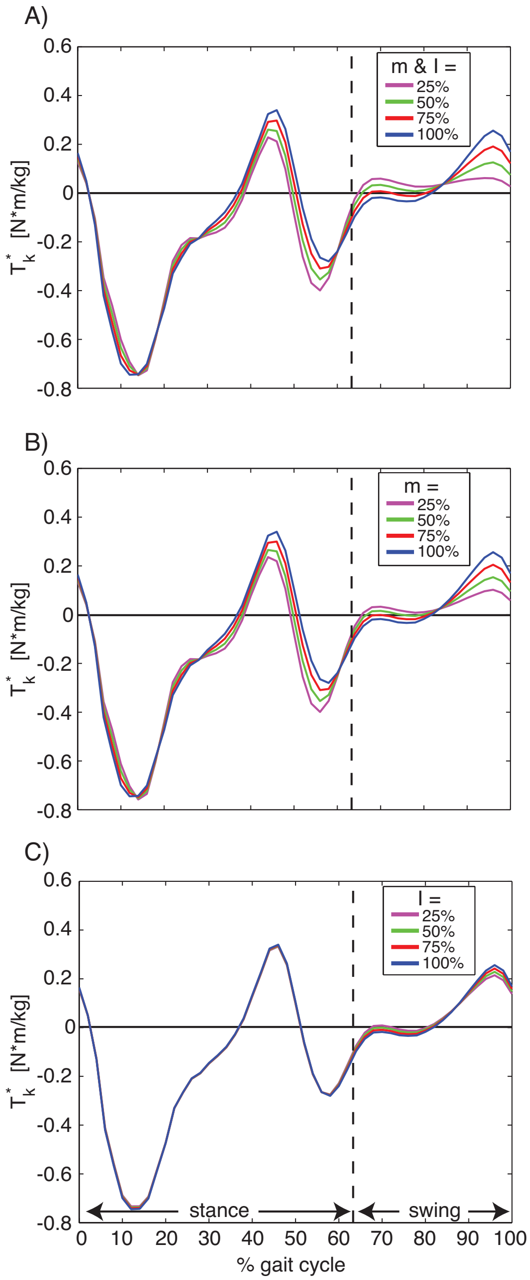

Fig. 3. Effects of altering prosthesis inertial properties on prosthetic knee moment. Symbol T_k^* designates knee moment normalized to body mass. Positive T_k^* values correspond to flexion moment and the negative T_k^* values correspond to extension moment. Alterations in inertial parameters apply to all leg segments and are specified as percentages of able-bodied values (e.g., "25%" indicates that specified inertial parameters for upper leg, lower leg, and foot are all scaled to 25% of corresponding able-bodied values). A) Masses and moments of inertia of all leg segments altered. B) Masses of all leg segments altered, but moments of inertia held constant at 100%. C) Moments of inertia of all leg segments (about corresponding COM) altered, but masses held constant at 100%.

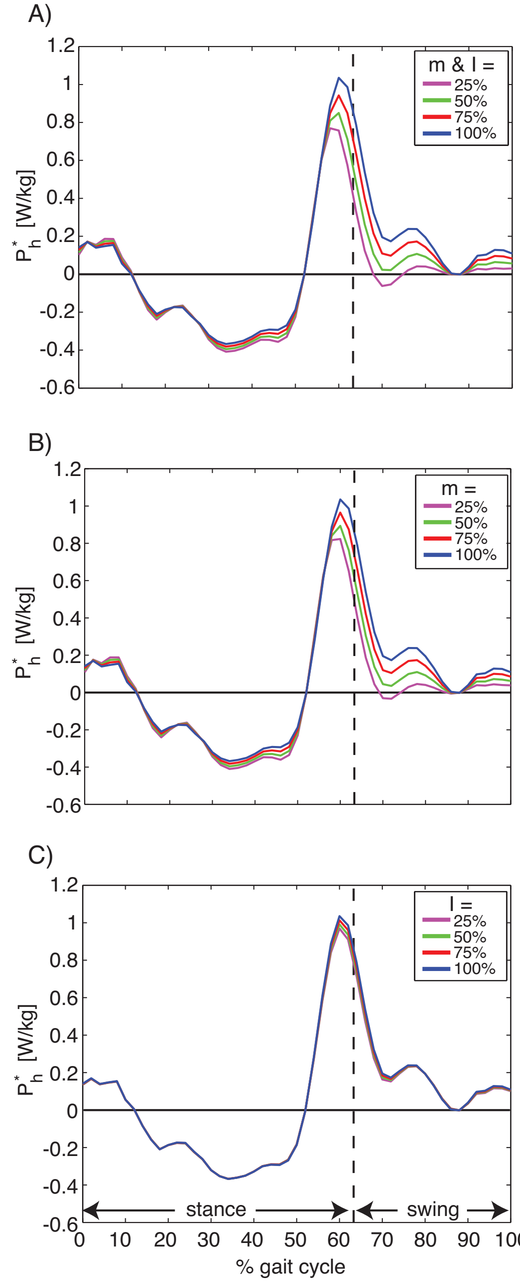

Fig. 4. Effects of altering prosthesis inertial properties on hip power. Symbol P_h^* designates hip power normalized to body mass. Alterations in inertial parameters apply to all leg segments and are specified as percentages of able-bodied values. A) Masses and moments of inertia of all leg segments altered. B) Masses of all leg segments altered, but moments of inertia held constant at 100%. C) Moments of inertia of all leg segments (about corresponding COM) altered, but masses held constant at 100%.

on knee extension moment during late stance, decreasing its peak magnitude by no more than 2%, and a small effect on knee moment during swing, decreasing its peak flexion magnitude by up to 16%. Thus, masses had a major effect on knee moment during the gait cycle, whereas moments of inertia did not.

B. Effects of prosthesis mass and moment of inertia on hip energetics

The effects of prosthesis inertial variations on hip power are illustrated in Fig. 4. Decreasing both masses and moments of inertia of all segments of the prosthetic leg had a large effect on hip power, reducing peak hip power during stance by up to 26% and average hip power during swing by up to 74% (Fig. 4A). Decreasing masses and holding moments of inertia constant had a similar effect, reducing peak hip power

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during stance by up to 20% and average hip power during swing by up to 66% (Fig. 4B). Decreasing moments of inertia and holding masses constant had a small effect on hip power, reducing peak hip power during stance by no more than 6% and average hip power during swing by no more than 8% (Fig. 4C).

The effects of prosthesis inertial variations on absolute hip work align with the results for absolute hip power (Table II). Decreasing both masses and moments of inertia of all segments of the prosthetic leg reduced absolute hip work by up to 22%. Decreasing masses and holding moments of inertia constant reduced hip work by nearly as much, 19%. On the other hand, decreasing moments of inertia and holding masses constant reduced absolute hip work by no more than 4%. Collectively, the preceding results demonstrate that variations in mass had a considerable effect on hip power and absolute hip work, but variations in moments of inertia had a negligible effect.

Fig. 4 also shows that decreasing prosthesis mass primarily reduced absolute hip work during swing. Decreasing prosthesis mass lowered the peak magnitude of late-stance hip power relative to able-bodied values, resulting in decreased absolute hip work during late stance. Decreasing prosthesis mass increased the magnitudes of early-to-mid-stance hip power relative to able-bodied values, resulting in increased absolute hip work during this part of the gait cycle. The reduced absolute hip work during late-stance and the increased absolute hip work during early-to-mid-stance were of nearly equivalent value. Thus, decreasing prosthesis mass had a negligible effect on absolute hip work during stance. In fact, when prosthesis mass was reduced to 25% of its corresponding able-bodied value, 2% of the total reduction in absolute hip work occurred during stance and 98% occurred during swing. This is in alignment with the common notion that inertial properties affect energetics more during swing phase as compared to stance phase.

TABLE II

EFFECTS OF ALTERING PROSTHESIS INERTIAL PROPERTIES ON ABSOLUTE HIP WORK OVER THE GAIT CYCLE. SYMBOL W_h^* DENOTES ABSOLUTE HIP WORK NORMALIZED TO ABSOLUTE HIP WORK OF PROSTHETIC LEG WITH ABLE-BODIED INERTIAL PROPERTIES. ALTERATIONS IN INERTIAL PARAMETERS APPLY TO ALL LEG SEGMENTS AND ARE SPECIFIED AS PERCENTAGES OF ABLE-BODIED VALUES

	m	I	W_h^*
m altered	25%	100%	0.81
	50%	100%	0.87
	75%	100%	0.93
	100%	100%	1.00
I altered	100%	25%	0.96
	100%	50%	0.97
	100%	75%	0.99
	100%	100%	1.00
m & I altered	25%	25%	0.78
	50%	50%	0.84
	75%	75%	0.92
	100%	100%	1.00

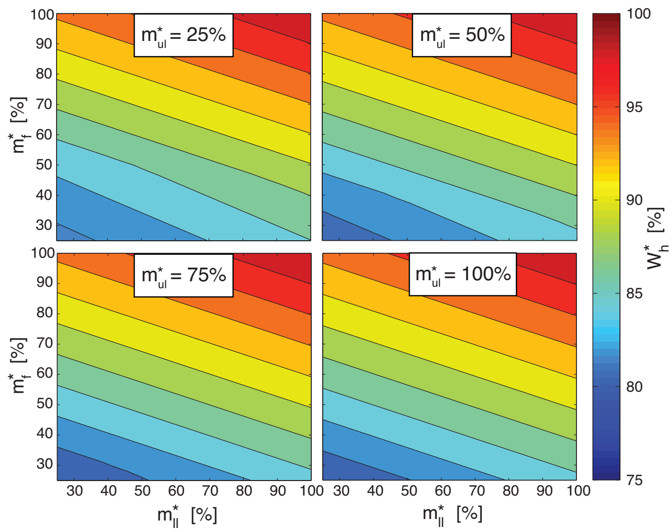

Fig. 5. Effects of lower leg mass and foot mass on absolute hip work for various values of upper leg mass. Symbols m_{ul}^*, m_{ll}^*, and m_f^* designate upper leg mass, lower leg mass, and foot mass normalized to corresponding able-bodied values. Symbol W_h^* is absolute hip work normalized to absolute hip work of prosthetic leg with able-bodied inertial properties. Graphs are presented for m_{ul}^* values of 25%, 50%, 75%, and 100%.

C. Effects of prosthesis segmental masses on absolute hip work

The effects of varying the masses of particular segments on absolute hip work are illustrated in Fig. 5. When upper leg mass, lower leg mass, and foot mass were all decreased to 25% of their corresponding able-bodied values, absolute hip work was reduced by 22%. Upper leg mass had little effect on absolute hip work. However, decreasing lower leg mass and foot mass to 25% of their corresponding able-bodied values reduced absolute hip work by 20-22%. Furthermore, the slope of the contour lines in Fig. 5 have a minimum magnitude of approximately $\frac{1}{3}$, indicating that absolute hip work was three times more sensitive to foot mass than to lower leg mass. We also observed that the effect of varying the masses of each of the segments on absolute hip work was almost linear.

IV. DISCUSSION

A. Analysis of major findings

1) Prosthetic knee moment: Varying prosthesis inertial properties was found to have a considerable effect on the prosthetic knee moment that would be required for an above-knee prosthesis user to walk with able-bodied kinematics, maximally altering the magnitude of moment by approximately 40-80% throughout the gait cycle. Comparison of our work with experimental studies examining the effects of prosthesis alterations on joint kinetics and energetics of above-knee amputees [41], [42], [43], [26] is challenging, as these studies typically applied mass perturbations to the prostheses of amputees that do not walk with able-bodied kinematics. However, the results agree with the theoretical

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study of Srinivasan [27], who evaluated the effects of mass added or removed at various locations along the shank of a below-knee prosthesis on total knee moment cost (i.e., the integral of the absolute value of knee moment over the gait cycle) for an anthropomorphic forward dynamic model of a transtibial amputee. Srinivasan's study was modeled for a below-knee prosthesis and no equivalent study for an above-knee prosthesis was found in literature. Srinivasan determined that as prosthesis mass is removed near the able-bodied COM of the lower leg, total knee moment cost during swing decreases by a large percentage, whereas total joint moment cost over the entire gait cycle decreases by a smaller percentage. These trends agree with those that can be visually assessed from Fig. 3.

For researchers and designers, our results signify that altering the mass or mass distribution of a prosthesis will considerably affect the prosthetic knee moment required for walking with able-bodied kinematics. Thus, when designing a prosthetic knee, the adjusted knee moment as a function of mass should be considered, as opposed to reproducing able-bodied knee moments. By parametrically modeling adjusted knee moments, mechanical components (such as springs, dampers, and actuators) can be selected to accurately replicate them. This conclusion is supported by the work of Sup et al. [20], which presented a powered robotic prosthetic knee with components optimized to reproduce able-bodied knee moment. The authors found that component parameters preferred by users of their knee were different from theoretically optimized parameters, and they proposed that inertial differences between users and able-bodied humans may have been the cause. Moreover, this conclusion holds even greater importance for users of low-cost, passive knees designed for the developing world. Many microprocessor controlled active knees can detect and compensate for undesired kinematics as a user walks [44], [12], [45]. However, low-cost, passive knees typically have components that have a very limited ability to adapt and change properties (e.g., spring and damping coefficients) during gait. These components must be selected and optimized correctly in order to enable desired kinematics, which is the focus of our ongoing work based on the results from this paper [46].

One aspect of our results may be counterintuitive. Although decreasing prosthesis mass generally reduced the knee moment required for able-bodied kinematics over the gait cycle, it actually increased the knee extension moment required in late stance. This phenomenon is a common occurrence for multi-rigid-body systems. For example, decreasing the mass of the foot will decrease the plantar-flexion moment required at the ankle during late stance. Because the reaction moment caused by the ankle plantar-flexion on the shank acts in the same direction as the extension moment applied by the knee at this point in gait, decreasing ankle plantar-flexion moment will increase the knee extension moment required to achieve the same total moment acting on the shank.

The results of this paper also demonstrated that varying prosthesis mass and moment of inertia together or varying prosthesis mass alone had a considerably greater effect on knee moment than varying moment of inertia alone. No other experimental or theoretical studies were found that evaluated the effects of independently altering mass and moment of inertia on the prosthetic knee moment required for able-bodied kinematics. For designers and prosthetists, these results signify that changing the density of material for a given prosthesis segment or adding mass at the COM will considerably affect the prosthetic knee moment required for able-bodied kinematics, whereas redistributing the material about the COM of the segment will not.

2) Hip energetics: Decreasing prosthesis inertial properties was found to have a major effect on the hip power and absolute hip work predicted by our model for an above-knee prosthesis user to walk with able-bodied kinematics, maximally reducing average hip power during swing by approximately 70% and absolute hip work over the gait cycle by approximately 20%. These results agree with other theoretical work reporting that mass perturbations applied at various locations along the shank-foot of an above-knee prosthesis cause hip power and absolute hip work to change considerably [47]. Numerical values could not be compared because the cited study considered adding and removing mass at the combined COM of the shank and foot while simultaneously altering moments of inertia in a non-proportional manner, whereas the present study considered decreasing masses of the segments separately and altered moments of inertia in proportion to masses. Our results also align with the theoretical work of Srinivasan [27], who evaluated the effects of mass added or removed at various locations along the shank of a below-knee prosthesis on total joint power cost (a quantity proportional to absolute joint work, but summed over multiple joints) for an anthropomorphic forward dynamic model of a transtibial amputee. Srinivasan determined that, for an optimal prosthesis alignment, removing mass near the COM of a prosthesis significantly decreased total joint power cost. Numerical values could not be compared because the present study calculated absolute joint work for the hip alone.

Our results also demonstrate that decreasing prosthesis mass and moment of inertia together or decreasing prosthesis mass alone cause a much larger reduction in hip power and absolute hip work than decreasing moment of inertia alone. Another theoretical study that independently varied mass and moment of inertia and computed hip energetic parameters could not be found for comparison.

In this study, reductions in the absolute hip work resulting from decreasing prosthetic mass occurred almost entirely during swing. This result can be readily explained by considering that the kinetic energy of any given segment of the prosthetic leg is equal to $\frac{1}{2}m|\overrightarrow{\dot{x}}|^2 + \frac{1}{2}I|\overrightarrow{\dot{\theta}}|^2$, where m is the mass of the segment, $\overrightarrow{\dot{x}}$ is the velocity of the COM, I is the moment of inertia about the COM, and $\overrightarrow{\dot{\theta}}$ is the angular velocity of the segment. Thus, decreasing prosthesis mass will reduce the kinetic energies of the segments. The kinetic energies themselves are significantly greater during swing than in stance [48]. Because the total change in the kinetic energies of all segments during a given phase of gait is equal to the sum of joint work and gravitational work during that phase, changes in joint work can be expected to be much greater

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during swing than in stance.

Our results also demonstrate that decreasing the masses of distal segments causes a much greater reduction in the absolute hip work required for able-bodied kinematics than decreasing the masses of proximal segments. Decreasing upper leg mass had a negligible effect on absolute hip work, and decreasing foot mass caused a threefold greater reduction in absolute hip work than decreasing lower leg mass.

For designers and prosthetists, these results quantitatively support and parametrically model a strategy that has been employed for decades: decreasing prosthesis mass is an effective means of reducing the mechanical energy of walking required for above-knee prosthesis users. Since jointly decreasing prosthesis mass and moment of inertia (physically, decreasing material density of the prosthesis segments) and decreasing prosthesis mass alone (physically, removing mass at the COM of each of the segments) were both found to cause considerable reductions in absolute hip work, decreasing material density or removing mass could both be feasible strategies for reducing mechanical energy expenditure. Since decreasing moments of inertia alone (physically, moving the mass of each prosthesis segment closer to its respective COM) did not considerably reduce absolute hip work, altering the mass distributions of the segments in this manner would not be a feasible strategy. In addition, since decreasing foot mass caused the largest reduction in absolute hip work, designing or selecting a lightweight prosthetic foot could be particularly effective at reducing mechanical energy expenditure.

Although the relationship between mechanical energy expenditure and overall metabolic energy expenditure for amputees has been debated [22], [43], [23], [49], it has been shown that metabolic cost is proportional to mechanical work in isolated muscles [50], [51], [52]. Thus, the findings presented here suggest that decreasing prosthesis mass in order to reduce mechanical energy expenditure at the hip could also reduce metabolic cost at the hip. Because transfemoral amputees suffer muscle cleavage, resulting in non-ideal muscle geometry of the hip muscles [53], decreasing prosthesis mass could be an effective strategy for reducing metabolic cost in a severely weakened area. This conclusion is particularly relevant for users of low-cost, passive prosthetic knees for developing countries, as these knees typically weigh in the same range as advanced microprocessor controlled and powered prostheses [54], [55], [45] and also lack the ability of recent active prostheses to generate positive mechanical work [45]. Designers and prosthetists can use the parametric illustration in Fig. 5 to determine the extent to which reducing the mass of one or more segments of a prosthesis could reduce absolute hip work, and thus lower metabolic cost.

B. Limitations of the study

Because of the lack of complete gait data sets in the literature for a large population of able-bodied adults, kinematic and kinetic data from multiple sources had to be estimated and combined. Joint angles were acquired from [30], and hip position, trunk angle, and COP location were extrapolated from other sources. In addition, the GRF distribution between legs during the double support phases was approximated. Although the joint moments calculated for able-bodied inertial properties closely matched able-bodied joint moments, the accuracy of the model could be improved through extensive data collection and more accurate approximations of GRF during double support (e.g. [56]).

In addition, since the prosthetic knee moment and hip energetics for able-bodied kinematics were calculated and able-bodied kinematics were applied to the model, it was assumed that the above-knee prosthesis provided the ankle moment necessary for able-bodied foot kinematics. Since the net energy produced at the ankle over the gait cycle was positive for all prosthesis inertial configurations examined [57], [30], the prosthesis could only enable able-bodied foot kinematics if the ankle joint provided sufficient power. McNealy [58] and Seroussi [59] experimentally quantified the considerable difference in ankle power between transfemoral amputees (wearing passive foot prostheses) and able-bodied subjects. Recent ankle-foot prostheses may be able to generate required power for able-bodied kinematics [60], [61]. However, low-cost, powered ankles for developing countries have not yet been developed. Optimizing the design of passive prosthetic ankles that store and return energy during a step, to behave as close to physiological as possible, is still an active area of research [62], [63], [64], [65].

We recognize that in a clinical setting, many factors beyond gait kinematics and kinetics, such as stability and socket comfort, must be considered when designing and fitting prosthetic limbs. We also recognize that most amputees walk with kinematics and kinetics that deviate from able-bodied gait. We used able-bodied gait as a performance target that amputees will hopefully attain in the future, using improved versions of both active and low-cost prosthetic limbs. In the near future, lightweight prostheses could be tuned for adjusted joint moments, as described in this study, in order to replicate able-bodied kinematics. The methods presented in this paper could also be used to investigate how inertial properties of prosthetic leg segments affect kinetics of alternate gaits.

Our future work will include measuring the differences between kinematics and kinetics predicted using our model, and those collected in clinical tests with amputees using prosthetic knees with tuned inertial properties and mechanical elements to produce desired joint moments [66], [67]. These tests will help refine our model for use in both clinical and research applications.

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