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Piloting a Novel Computational Framework for Identifying Prosthesis-Specific Contributions to Gait Deviations

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ABSTRACT

This paper introduces a novel computational framework for evaluating above-knee prostheses, addressing a major challenge in gait deviation studies: distinguishing between prosthesis-specific and patient-specific contributions to gait deviations. This innovative approach utilizes three separate computational models to quantify the changes in gait dynamics necessary to achieve a set of ideal gait kinematics across different prosthesis designs. The pilot study presented here employs a simple two-dimensional swing-phase model to conceptually demonstrate how the outcomes of this three-model framework can assess the extent to which prosthesis design impacts a user's ability to replicate the dynamics of able-bodied gait. Furthermore, this framework offers potential for optimizing passive prosthetic devices for individual patients, thereby reducing the need for real-life experiments, clinic visits, and overcoming rehabilitation challenges.

1 | Introduction

Irregular walking patterns are often the result of prosthetic users looking to alleviate pain, increase speed, or improve stability [1, 2], especially when adapting to a new device. Furthermore, inadequate devices can be accompanied by mental blocks, leading to asymmetric gait, as the user relies more on the unaffected limb for support [3, 4]. Ongoing research has significantly improved prosthetic performance, with innovations such as smart and adaptive devices and 3D printed prostheses offering cost-effective DIY options [5–8]. Research groups have also explored the use of virtual environment processes to help prosthetists and technicians design and test knee prosthetic devices and their components [9–11].

Traditional gait evaluation relies on motion capture to observe and quantify kinematic trajectories, external loads, energy expenditure, and spatio-temporal gait parameters [3, 4, 12–14]. Kinematics can be combined with computational models to estimate joint loading and musculoskeletal dynamics [15–18]. Evaluating the effects of amputation and prosthesis design typically involves comparing prosthesis users to nonusers, making it challenging to distinguish between device effects and user-specific psychological and physiological factors, such as fear of falling, pain, fatigue, reduced muscle strength, and amputation-caused decreases muscle lever arm on a residual limb. The inherent heterogeneity in population demographics and amputation-related comorbidities adds complexity to identifying underlying causes. Mitigating confounding effects by studying gait dynamics for a single individual pre- and post-amputation or with multiple devices is possible [19, 20] but comes with the challenges of training periods and high prosthetic device costs.

This study presents a new computational framework designed to overcome limitations in identifying the distinct impacts of prosthetic knee design and configuration on gait dynamics. This framework employs three two-dimensional gait models to examine joint kinematics and kinetics; however, more sophisticated musculoskeletal models can be employed. Historically, musculoskeletal models of varying complexity have been applied to evaluate gait dynamics in prostheses users. The novelty of this research stems from the unique comparisons made between the dynamic outputs of each model. These comparisons are intended to pinpoint and comprehend the sources of gait deviations, determining whether they stem from the users or the prosthetic devices [21–26].

To highlight the conceptual validity of the framework, this study focused on evaluating the kinematics and kinetics of the joints in the sagittal plane during the swing phase of the prosthesis side limb. The three models used reference kinematic data from the gait of able-bodied individuals [27], generating required joint moments as output. The moments generated were then compared with identify how specific parameters contribute to gait deviations. The three-model framework also offers the advantage of predicting how individual design parameters of a prosthetic device can be adjusted to achieve the desired gait outcomes.

2 | Novel Approach and Framework

2.1 | Models

Our approach for evaluating prosthesis performance involves comparing outputs from three forward-inverse dynamic models, each driven by the same reference gait kinematics. While any common reference kinematics can be used, we chose able-bodied kinematics for this pilot study. However, it is important to recognize that the ideal kinematics for prosthesis-users may vary among individuals and accordingly other reference kinematics can be used [28]. By using the same reference kinematics as input to three different models, our framework effectively isolates the contributions of both the prosthetic device and the user to gait deviations, guided by three key questions:

1. What joint kinetics produce typical gait kinematics in able-bodied individuals?
2. How do changes in inertial properties resulting from the prosthetic device influence gait dynamics?
3. What joint kinetics are required to achieve able-bodied kinematics while using a specific prosthesis configuration?

These three questions lead to three distinct models in our framework: the Able-Bodied Model, the Ideal Prosthesis Model, and the Full Prosthesis Model. These models provide the joint kinetics outputs O1, O2, and O3, respectively, required for achieving able-bodied gait as conceptually depicted in Figure 1a The results of this paper were obtained by models developed in the SIMSCAPE MULTIBODY environment in MATLAB SIMULINK (R2021a, The MathWorks, Natick, MA). Each model targets the prescribed reference kinematics [27] or is actuated by the prescribed joint torques. The anthropometric measurements for constructing the models (Table 1) were derived using measurements, body segment lengths, and mass estimates from [27]. Each model consists of three independently actuated linked segments. The linked segment method in the sagittal plane, commonly used for gait modeling, provides an appropriate level of complexity and accuracy [16, 21, 29, 30]. The three links in the model are connected by revolute joints that allow rotations in the sagittal plane. The first link represents the thigh, hinged at the hip, which is considered a fixed point in this preliminary work. The second link represents the shank, attached to the thigh via the knee joint. Finally, the foot is connected to the shank through the ankle joint, which completes the assembly.

The individual details of the three models are next described for a simple swing phase. Although the simplest models used here offer insight into how well-designed comparisons resolve the root causes of gait deviations, for real application, an appropriately sophisticated musculoskeletal models would need to be used in our framework.

2.1.1 | Able-Bodied Model

The Able-Bodied model represents the leg of an able-bodied individual. The pendulum links were approximated as cylinders of equal diameter following the [16, 21] approach. The target kinematics of the joints [27] were input into the model, and the inverse dynamics simulations were used to compute (1) the joint kinetics required to produce typical gait kinematics in able-bodied individuals. Figure 2 validates the model by comparing the output, O1, with the values from [27]. Although the calculated hip and knee torques generally match the published values, noticeable differences are observed, particularly in the ankle torque during the late swing phase. These differences are expected due to the simplifying assumptions we made in the model and differences in the inertial properties of each body segment. A forward simulation confirmed that these torque estimates produced the original target kinematics, ensuring the internal consistency of the model.

2.1.2 | Ideal Prosthesis Model

The Ideal Prosthesis model is an adaptation of the Able-Bodied model that recalibrates segment inertia to reflect the mass distribution changes caused by replacing a natural limb with a prosthesis. Table 1 provides detailed information on these inertia properties. Furtherore, in this model, the shank segment is divided into a prosthetic knee and a cylindrical pylon. The prosthetic knee, connected to the thigh through the knee joint, rotates at one end and is rigidly connected to the pylon on the other. The foot segment is connected to the pylon through the ankle joint, and rotation is not allowed due to the minimal impact of the ankle motion during swing. Future work will incorporate ankle motion and more refined models with a stance phase. The length of the prosthetic knee is taken from the manufacturer's datasheet, while the lengths of the cylinder stud and foot are approximated to mimic the physique of the able body model. The masses of the prosthetic segments were approximated to represent the masses of the physical prosthetic components. The masses of the knee and foot were obtained from the manufacturer's datasheets [31]. The cylinder represents a standard 30 mm high-impact aluminum pylon.

Unlike the Full Prosthesis model described later, this model does not incorporate the complete mechanism of the device. Instead,

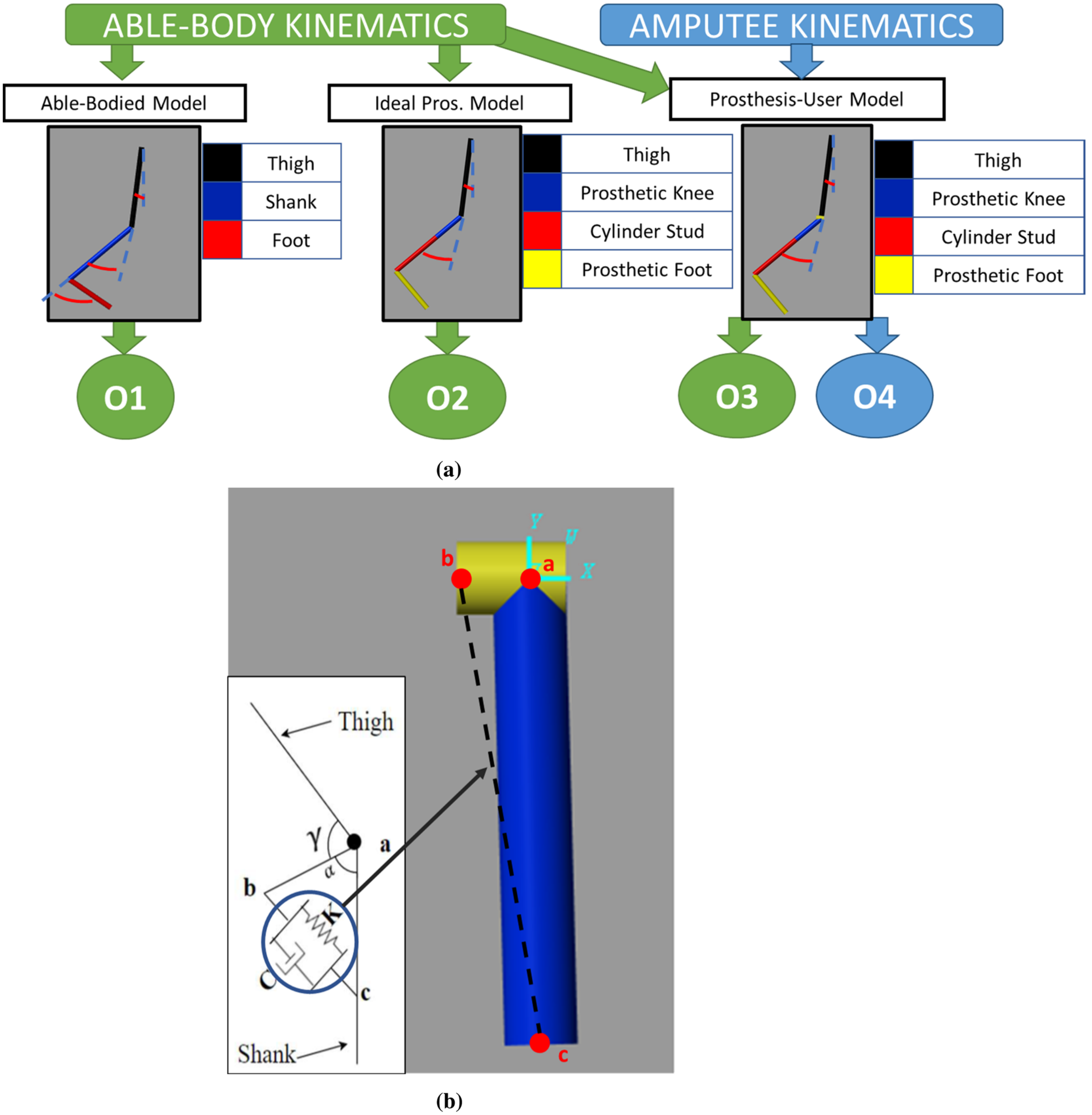

FIGURE 1 | (a) Three computational models starting from the left, Able-Bodied, Ideal Prosthesis, and Full Prosthesis, using different kinematics reference inputs (able-body and amputee) to produce the gait's dynamics. For the preliminary results of the study, the models solely represent the leg of an individual during the swing phase and use data from [27] as the reference input. (b) Prosthetic Knee Model designed in SIMSCAPE MULTIBODY and its description, the yellow segment is fixed to the thigh, and the blue segment rotates about point A. Points B and C are connected by a spring and damper, the natural position of the knee has the two segments perpendicular to each other.

the knee is treated as an ideal joint capable of generating the torques necessary to achieve the target kinematics. This model approximates an individual using a prosthetic knee joint that perfectly reproduces the dynamics of an intact knee. Traditionally, prosthetic devices have been designed to match the moments exhibited by able-bodied individuals [21]; however, the difference in inertial properties between able-bodied and prosthetic limbs should result in different joint moments. The model used inverse dynamics to compute the torque values required to achieve able-bodied kinematics with altered mass distributions. The results are saved for comparison as Output 2 (Figure 1a), and they inform us on (2) how changes in inertial properties resulting from the prosthetic device influence gait dynamics.

2.1.3 | Full Prosthesis Model

The Full Prosthesis model was built by replacing the placeholder prosthetic knee with a computational model of the prosthetic knee (Figure 1b). In this study, the Mauch S-N-S

TABLE 1 | Models segments' dimensions and inertial properties.

Model	Segment	Length (m)	C.o.M. (m)	Mass (kg)	Moment of inertia ($kg \times m^2$)
Able-Body	Thigh	0.394	0.170	5.670	0.092
	Shank	0.394	0.170	2.637	0.037
	Foot	0.243	0.122	0.822	0.011
Ideal Prosthesis	Thigh	0.394	0.170	5.670	0.092[a]
	Knee	0.160	0.080	1.140	5.75E-3
	Pylon	0.233	0.117	0.250	1.14E-3
	Foot	0.243	0.122	0.960	4.77E-3
Full Prosthesis	Thigh	0.394	0.170	5.670	0.092
	Knee (ac)	0.160	0.080	1.140	2.36E-3
	Pylon	0.233	0.117	0.250	1.14E-3
	Foot	0.243	0.122	0.960	4.77E-3

[a]The center of mass position is given with respect to the proximal joint. The segments' masses and lengths are based on anthropometric data estimates and measurements from the able-body walking trial in [27]. Mass = 56.7 kg, Height = 1.6 m.

prosthetic knee was modeled. Although future work will explore various prosthetic knee systems, the S-N-S knee was chosen for its relative simplicity and widespread use to illustrate the proposed modeling framework. The simulated prosthesis was modeled in a previous study [32]. Figure 1b shows the Mauch S-N-S model, comprising a fixed segment (ab) linked to the thigh and a second segment (ac) connected to the top of the first segment by a revolute knee joint. A combined spring and damper system connects the ends of the segments, acting as a substitute for the Mauch S-N-S knee cylinder (dashed black line in Figure 1b). The system is based on the two-phase model from [32], which is governed by the following equation:

$$F = (1 - S_3)(c_1\dot{x} + k_1x + f_1) + S_3(c_2\dot{x} + k_2x + f_2) \quad (1)$$

where x and $\dot{x}$ are the position and velocity of the piston, c_1, k_1, f_1, c_2, k_2, and f_2 are coefficients dependent on the phase, and S_3 is a variable responsible for the phase switch. As indicated in [32], Phase 2 is the low-force section (non-stance flexion phase of the gait); hence, during the swing phase, the equation can be reduced to

$$F = c_2\dot{x} + k_2x + f_2 \quad (2)$$

Initial values of the spring and damping coefficients were obtained from [32] to approximate the performance of a standard S-N-S knee configuration.

These values were later modified during the optimization part of this study to identify the configuration that minimizes kinematic deviations from able-bodied gait. Segment ac accounts for most of the prosthetic knee mass from [31]. The model does not include a prosthetic socket and is assumed to be rigidly connected to the distal end of the thigh. The natural length, L_0, of the system is given by the distance between the ends b and c (Figure 1b) when the knee flexion angle is 0:

$$L_0^{\,2} = ab^2 + ac^2 \quad (3)$$

where $ab = 0.0247$ m and $ac = 0.1603$ m.

Unlike the Able-Bodied and Ideal models, the Full Prosthesis model knee joint does not provide active actuation at the knee joint. Only the hip joint can generate positive work to provide the torque required to produce an able-bodied hip trajectory. The dynamic outputs of interest for the Full Prosthesis model (Output 3) are the hip moment required to follow the desired hip trajectory, the knee moment due to motion and the spring-damper system, and the kinematics of each joint. Ultimately, the model produces (3) the joint kinetics required to achieve able-bodied kinematics while using a specific configuration of the S-N-S knee. An eventual set of outputs for the model, shown in Figure 1a as Output 4, is (4) the joint kinetics actually exhibited by individuals using a specific prosthesis configuration. Although beyond the current scope, this set of outputs is derived from kinematic data of amputee subjects to finalize the new framework for assessing deviations. Ideally, these kinematics would be obtained from patients matching the mass, height, age, and sex of able-bodied subjects. They will be applied to models scaled for the users incorporated into the worn prosthetic device. A summary of the models, outputs, and descriptions/goals is presented in Table 2.

2.2 | Comparisons

Our strategy for assessing gait deviations in prosthesis users is grounded in the comparisons between the kinetic and kinematic generated outputs (Figure 1a) of each model. These comparisons serve to characterize and isolate prostheses and user-specific contributions to gait deviations.

- O1 versus O2: Evaluate the impact of the inertial changes due to the prosthetic device
- O2 versus O3: Evaluate/isolate the impact of a specific device on gait

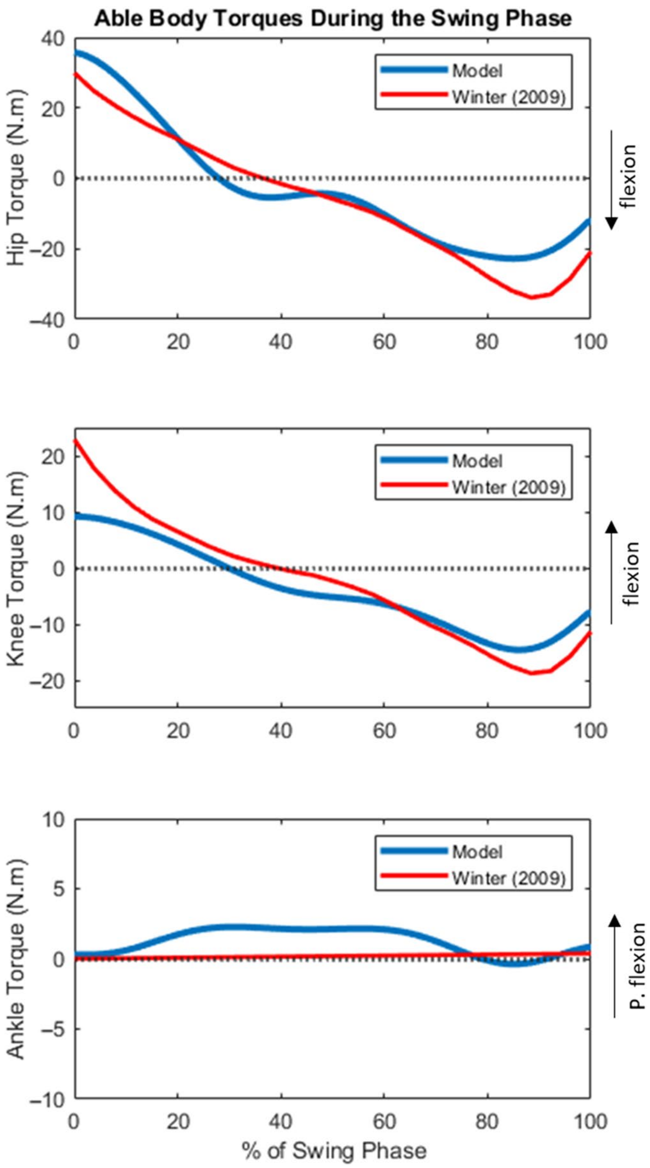

FIGURE 2 | Torques required by able-body model in comparison with data from [27] during the swing phase. O1: Output from the Able-Bodied Model.

TABLE 2 | List of the models, their outputs, and intended goals.

Models	Output and description/goal
I—Able-Bodied	O1: The joint kinetics required to produce typical gait kinematics in able-bodied individuals
II—Ideal Prosthesis	O2: How changes to the inertial properties of a healthy limb affect gait dynamics
III—Full Prosthesis	O3: The joint kinetics required to produce able-bodied kinematics while using a specific prosthesis configuration
	O4: The joint kinetics actually exhibited by individuals using a specific prosthesis configuration (not in current scope)

The significance of these comparisons hinges on the idea that driving multiple models with identical inputs and observing varying outputs indicates that the underlying differences between these models are the source of disagreements. The ideal prosthesis model and its output (Output 2) are used primarily as an intermediate step for our comparisons. While comparing Output 1 and 3 can be valuable, it does not consider changes in inertia between the able-bodied individual and the prosthesis user. Therefore, the comparison between outputs 1 and 2 isolates the effects of the inertial properties of the limb on gait dynamics [21].

The able-bodied kinematics used were chosen to ensure that each model's results represented the joint kinetics required to execute the target motion. When applied to the Ideal Prosthesis model, it represents what an amputee is required to do. When applied to the Full Prosthesis model, it depicts what an amputee would do to achieve the desired motion with a specific prosthetic device. The prosthetic devices can be evaluated by comparing Outputs 2 and 3 (Figure 1a), and the differences between various devices can be assessed by comparing the outputs (O3) of the Full Prosthesis model equipped with different knee mechanisms.

The first two comparisons are valuable for understanding the contribution of a specific device to gait deviations, which is the main focus of the current study. As for the other source of deviations, isolating the patient's contribution involves providing two types of input to the full prosthesis model: Able-Bodied and Prosthesis-User kinematics. Variations in outputs result from differences in target input, revealing the specific contribution of the knee model or patient to the deviations. These comparisons will be the subject of subsequent studies.

- O1 versus O4: Traditional evaluation, comparing the gait dynamics of an able-body individual and amputee
- O3 versus O4: Evaluate/isolate the impact of the patient's contribution on the gait

2.3 | Parameter Estimation/Optimization

Our new framework offers the possibility of modifying and optimizing the design of prosthetic knees. Thanks to the modular approach, a specific device can be incorporated within the models, tested, and its specifications changed. Parameter searches were performed on the Mauch S-N-S Prosthetic Knee with the primary goal of finding a combination of spring and damping constants for the prosthetic knee that would best match kinetic or kinematic targets. An iterative process was utilized with our models in the SIMULINK parameter estimator tool, first using a nonlinear least-squares approach. The cost function is the sum square error and the initial values for stiffness and damping are from [32]. Three sets of initial values over an extensive range were tested and converged to the same values for the parameters. A pattern search followed the first round to determine the optimal coefficients.

The full prosthesis model includes a fully passive prosthetic knee designed according to the geometric description and receives no actuation. The only actuation provided to the model is from the hip joint. For optimization, the hip is driven by the healthy hip joint kinematics and the software then iterates and uses the healthy knee joint kinematics as targets to find the desirable spring and damping constants. This shows the benefits of the new approach we propose, as opposed to a real-life experiment and lab observation process that would require considerable effort on the patients and add physical and mental toll during tuning.

3 | Results and Discussion

When comparing O1 and O2, the gait dynamics changes noticeably due to differences in inertial properties (Figure 3). Both models show a similar trend in torques at the hip and knee joints, with the ideal prosthesis model exhibiting smaller magnitudes. This decrease is attributed to the lighter nature of prosthetic devices compared with limbs. Significant differences emerge around the 25%–40% and 70%–90% windows, corresponding to peak knee flexion and late-swing extension. The smaller mass of the Ideal Prosthesis results in less inertia, requiring additional hip extension torque during peak knee flexion and less torque during late-swing extension. These findings align with previous research [21], emphasizing that reducing mass and inertia in prosthetic devices significantly decreases knee moment and hip power needed during the swing.

When analyzing the impact of the knee components on gait within the full prosthesis model, simulations were performed by varying the stiffness and damping coefficients of the hydraulic knee joint to represent different configurations. These included a typical knee, a spring-dominant knee, and a viscous-dominant knee. Initial values were obtained from the Mauch S-N-S model in [32], the remaining spring and damping constants are listed in Table 3. Figure 4 illustrates that the spring-dominant knee configuration requires larger hip torque, than what is required by the other two configurations, to achieve the target kinematics, but cannot achieve full knee flexion. In contrast, the knee with higher damping parameters aligns closely with the able-bodied hip torque profile but exhibits incomplete knee extension at the end of the swing phase. These findings underscore the models' ability to predict the effects of design parameters on user gait dynamics. They offer valuable insights for customizing device configurations to minimize gait deviations for individual users, potentially helping patients and prosthetists during rehabilitation.

The spring stiffness and damping coefficients were optimized to minimize deviations in knee kinematics from able-bodied trajectories (angle and angular velocity, Figure 5). The hip torque required for the optimized prosthetic knee joint was compared with the able-bodied torques (O1) and the ideal model torques (O2). In Figure 3, similar torque trends are observed at the hip, with notable distinctions. Specifically, the

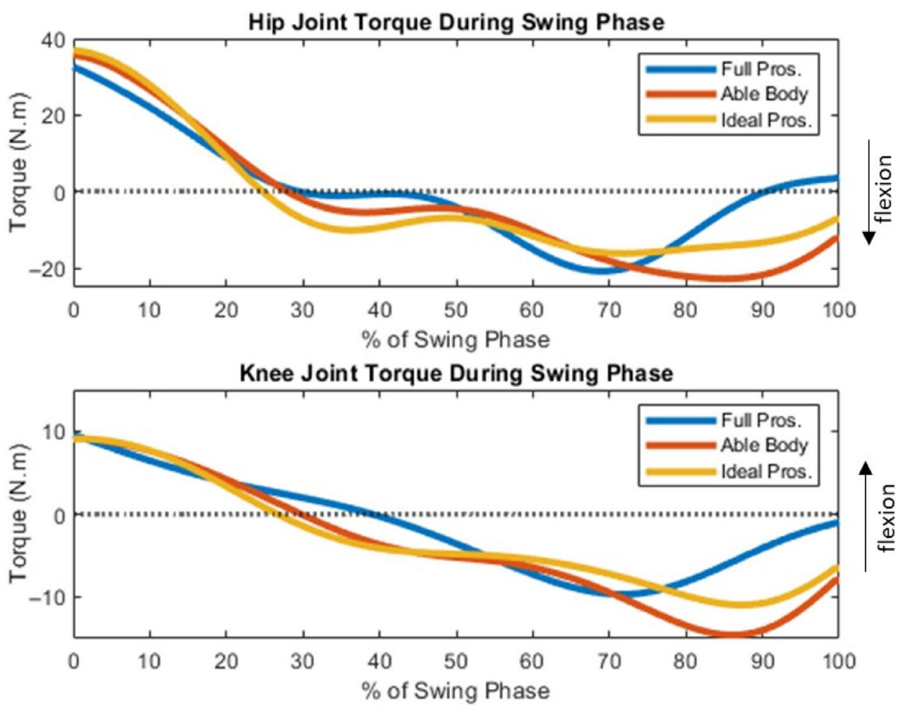

FIGURE 3 | Torques at the hip and knee versus time during the swing phase for the 3 models: Able body (O1), ideal Prosthesis (O2), and full Prosthesis after knee optimization (O3).

TABLE 3 | Spring and damping constants utilized for the simulations (Figure 4).

	From literature [32]	High stiffness	High damping	Optimized values
Spring (N/m)	14,116	42,348	0	7929.79
Damper (N.s/m)	1415.05	0	4245.1	3204.63

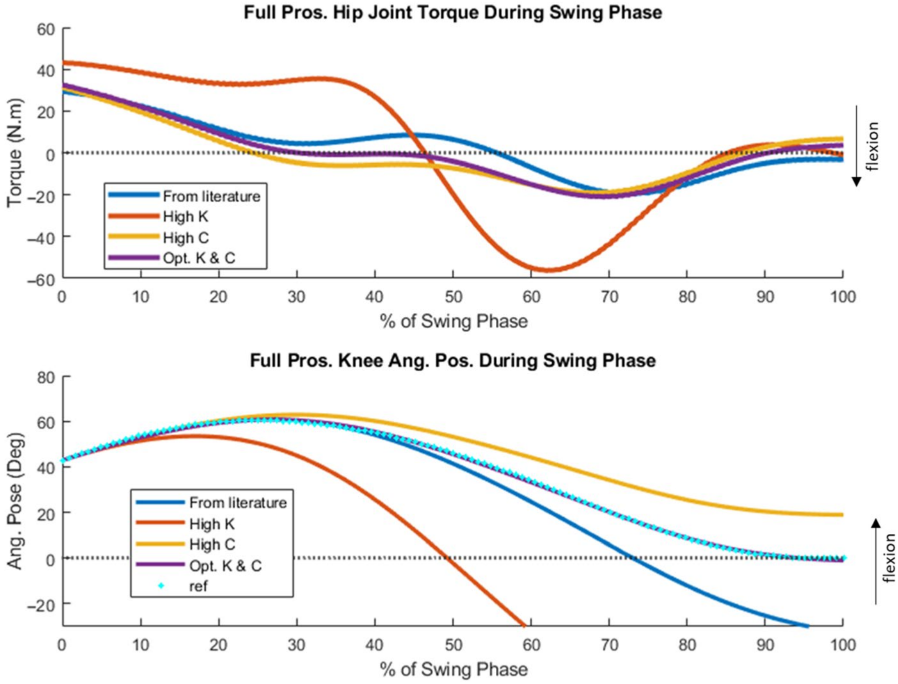

FIGURE 4 | Comparing the effects of various knee parameters on the gait dynamics of the Full Prosthesis model, particularly the hip torque and knee angular position, by changing the spring and damping constants. The results of the model with parameters from the literature [32] are represented by the curves in blue. The results of the models with the high stiffness and high damping are represented with the curves in orange and yellow respectively. The purple curves represent the results of the model equipped with the optimized knee, and the cyan curve (reference) shows the knee angular position of the able body used as a reference.

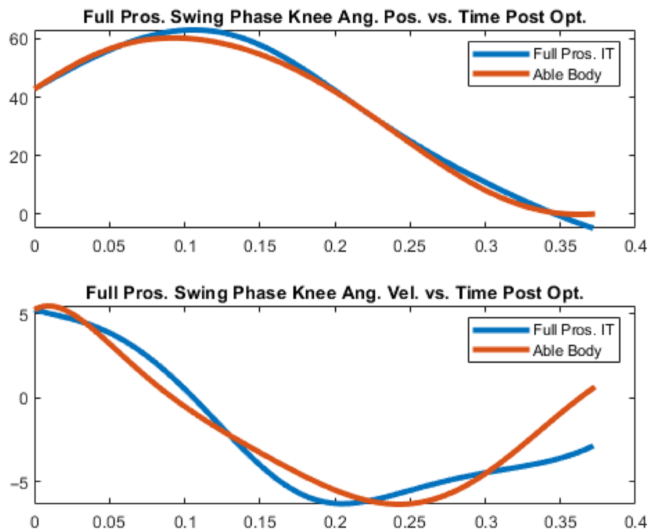

FIGURE 5 | Knee joint angular position and velocity versus time during swing phase post-optimization of the knee while driving the hip with able-body kinematics from [27].

optimized Full Prosthesis model (O3) displays an earlier peak extension torque in the late swing compared with the other two models. This early peak extension torque signifies the necessity for the hip to halt its forward motion when using a passive knee design, allowing ample time for the knee to reach full extension in preparation for heel strike. This outcome aligns with the smaller extension torques of the prosthetic knee (O3) in contrast to the actively actuated knees (O1 and O2).

Our preliminary results build a foundation to support our new approach to prosthetic gait evaluation, but it is understood that they primarily serve as concept validations. To complete and verify our approach for evaluating prosthetic gait, the fourth output, O4 (Figure 1a), is required for comparison. O4 will offer information on the potential strategies used by amputees to compensate for prosthesis limitations and/or patient-specific health factors affecting gait dynamics. Future work involves completing the gait cycle with more refined models, addressing assumptions about differences between thigh segments, weights of sockets, liners, connectors, and incorporating the prosthetic foot. To compare kinematics data of amputees and outputs from our prosthesis model, we plan to generate able-bodied data from published datasets [33], ensuring matching mass/height/age and gender with the amputees. Additionally, implementing three-dimensional models is necessary to capture deviations of prosthesis users outside the sagittal plane.

4 | Conclusion

Despite the relative simplicity of the two-dimensional models used, the changes in torque magnitudes between O1, O2, and O3, and the optimized prostheses parameters demonstrate the power of our three-model approach to isolate the contribution of the device to gait. Future work will build upon this strong foundation to obtain more clinically relevant insights by including three-dimensional modeling, both the stance and swing phases, and empirical data from prosthesis users for clinical validation. Ultimately, this framework will drastically reduce the time and cost inherent to traditional gait evaluation and will facilitate customized prosthesis designs and clinical rehabilitation strategies that are optimized to meet the needs of individual users.

Author Contributions

Conceptualization: M.L. and S.G. Methodology: M.L. Investigation: J.-E.N. Writing – original draft preparation: J.-E.N. and M.H.A. Writing – review and editing: M.L. and S.G. Supervision: M.L. and S.G.

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Ethics Statement

The authors have nothing to report.

Conflicts of Interest

The authors declare no conflicts of interest.

Data Availability Statement

Data sharing is not applicable to this article as no new data were created or analyzed in this study.

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