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Approach for gait analysis in persons with limb loss including residuum and prosthesis socket dynamics

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Abstract

Musculoskeletal modeling and marker-based motion capture techniques are commonly used to quantify the motions of body segments, and the forces acting on them during human gait. However, when these techniques are applied to analyze the gait of people with lower limb loss, the clinically relevant interaction between the residual limb and prosthesis socket is typically overlooked. It is known that there is considerable motion and loading at the residuum-socket interface, yet traditional gait analysis techniques do not account for these factors due to the inability to place tracking markers on the residual limb inside of the socket. In the present work, we used a global optimization technique and anatomical constraints to estimate the motion and loading at the residuum-socket interface as part of standard gait analysis procedures. We systematically evaluated a range of parameters related to the residuum-socket interface, such as the number of degrees of freedom, and determined the configuration that yields the best compromise between faithfully tracking experimental marker positions while yielding anatomically realistic residuum-socket kinematics and loads that agree with data from the literature. Application of the present model to gait analysis for people with lower limb loss will deepen our understanding of the biomechanics of walking with a prosthesis, which should facilitate the development of enhanced rehabilitation protocols and improved assistive devices.

KEYWORDS

inverse dynamics, inverse kinematics, prosthesis

1 | INTRODUCTION

Modern marker-based motion capture techniques combined with musculoskeletal modeling and computational analysis have become standard tools for studying normal and pathological gait biomechanics. However, current techniques are limited when used to analyze the biomechanics of gait in people with lower limb loss who use a prosthesis. There is considerable motion of the residual limb inside the socket,[1-5] but that motion and the corresponding loading are not captured in routine gait analysis due to the inability to place tracking markers on the residuum inside of the socket. In inverse dynamics studies (eg, Winter and Sienko[6]), motion at the residuum-socket interface will cause changes in the effective length of the single modeled segment running from the knee to the ankle by amounts that vary over the gait cycle, and differ among subjects. Conversely, in forward dynamics simulation studies (eg, Fey et al[7]) the residuum-socket interface is usually assumed to be rigid, despite the motion that is known to exist. The residuum-socket interface should be treated explicitly as another joint in the system, but this presents challenges when using traditional marker-based kinematic techniques. Here, we describe an approach for estimating gross motion and resultant loads at the residuum-socket interface as part of whole-body inverse kinematics (IK) and inverse dynamics (ID) analyses. Our perspective is that this approach will lead to a more comprehensive understanding of gait biomechanics in persons with lower-limb amputation who walk with a prosthesis and also provide direct insights into the mechanics of the residuum-socket interface.

Previous studies of gait biomechanics in persons who walk using a prosthesis have commonly used direct methods[8] for determining gait kinematics that effectively treats the residuum and prosthesis socket/pylon as a single segment. Thus, while the motion at the residuum-socket interface is present in the experimental data (eg, marker data), these motions and the corresponding forces remain obscured. To address this limitation, researchers have used various sensing technologies to directly record residuum-socket motion including ultrasonography,[5] radiography,[2,4,9,10] fluoroscopy,[11] and custom proximity sensors.[12] However, such methods are complex, require costly or highly specialized equipment, and, in some cases, expose the subject to potentially harmful radiation making these methods impractical for routine gait analyses.

Current marker-based IK methods estimate all underlying body segment positions and orientations by utilizing a global least-squares optimization, subject to anatomical joint constraints, as first described by Lu and O'Connor.[13] This approach provides the opportunity to compensate for the inability to place tracking markers on the residuum inside of the socket. The constraints imposed by a residuum-socket joint in a model directly representing the amputation, combined with tracking markers proximal to the joint (eg, thigh, pelvis) and distal to the joint (eg, socket, foot), could provide the basis for reconstructing the residuum-socket kinematics. We propose an approach for whole-body IK and ID analyses utilizing a least-squares global optimization method employing a 3-dimensional musculoskeletal model including a residuum-socket joint to determine kinematics and kinetics at this interface simultaneously with traditional joint-level biomechanical variables (eg, angles, moments).

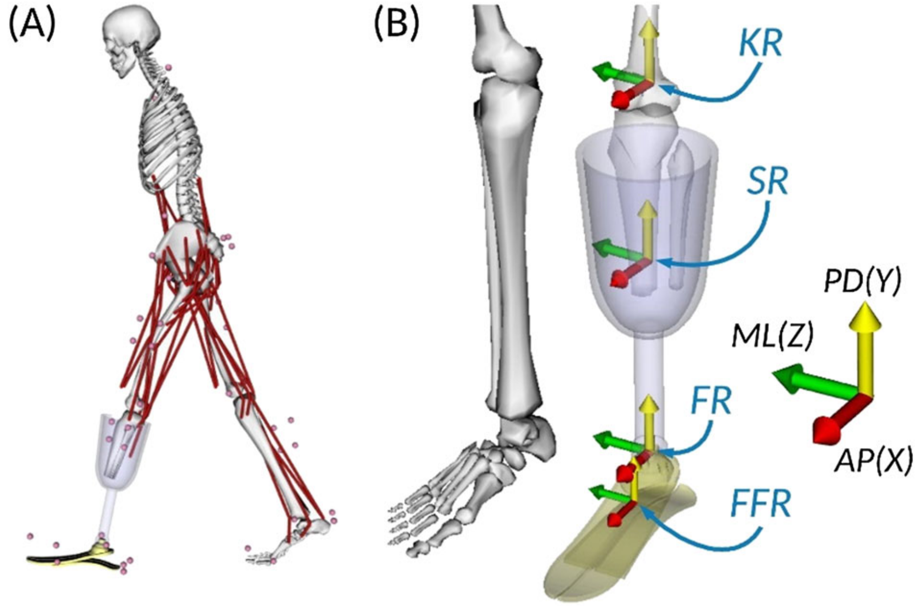
In this paper, a gait analysis model representing people with unilateral transtibial amputation was developed in OpenSim.[14,15] The joint between the residuum and prosthesis socket has maximally 6 DOFs. Each individual DOF can be constrained providing the residuum-socket joint with anywhere between zero and 6 DOFs. Experimental walking data for 3 subjects with unilateral transtibial amputations were used to demonstrate the utility of the model. A global least-squares approach was used to perform IK analyses with the residuum-socket model in different configurations followed by ID analyses. The effects of different DOFs in the residuum-socket joint on the IK marker tracking error were examined, and the resulting residuum-socket generalized motions and forces for each model configuration were compared for all 3 subjects.

2 | METHODS

2.1 | A generic musculoskeletal model with unilateral transtibial amputation

Generic models with left and right unilateral transtibial amputation (Figure 1A) were created by modifying the "gait2354" model provided with OpenSim.[14,15] Both versions of the model may be downloaded at https://simtk.org/projects/bkamputee_model. All bodies distal to the affected tibia were removed, and the affected tibia body segment (which includes the fibula) was transected by modifying mass, inertia, and the graphical depiction to represent an amputation at 50% of the limb length. All muscles distal to the amputation site in the generic model were removed except the biarticular gastrocnemius. The gastrocnemius was reattached arbitrarily to the proximal posterior tibia, maintaining a musculoskeletal model that would numerically satisfy OpenSim analysis requirements and could later be modified to represent a subject-specific muscle reattachment surgery. A generic socket body segment, including the pylon, was articulated with the transected tibia parent body via a reversed joint having 6 independently constrainable DOFs expressed in the socket reference frame SR (Figure 1B). DOFs included flexion/extension, abduction/adduction, axial rotation, pistoning (axial translation), anterior-posterior translation, and medial-lateral translation.

A model of an ankle-foot prosthesis was connected to the end of the pylon on the socket body segment at the foot reference frame (FR) (Figure 1B). This was initially placed in the socket segment frame such that the prosthetic foot was aligned with the unmodified model. Although some forward dynamics studies have incorporated prosthetic foot models containing many segments to approximate continuous flexion throughout the prosthetic foot,[7] the ankle-foot flexion defined in this model is lumped into a single DOF pin joint at the foot flexion reference frame (Figure 1B), as commonly seen in the literature.[16,17] A lumped flexion parameter has the advantage of being computationally efficient while



(C) **Socket-Limb Joint Configurations**

	Free Rotational Coordinates			Free Translational Coordinates		
	AP(X)	PD(Y)	ML(Z)	AP(X)	PD(Y)	ML(Z)
Rigid – Welded						
Flex - Flexion Only			•			
Pist – Pistoning Only					•	
Flex/Pist - Flexion and Pistoning			•		•	
4 DOF - Full Rotation with Pistoning	•	•	•		•	
6 DOF - Full Rotation and Full Translation	•	•	•	•	•	•

FIGURE 1 A, A generic model with transtibial amputation was created by modifying the gait2354 model provided by OpenSim. B, A closeup of the socket-limb model is shown with the reference frames used for each joint, and coordinates of the constrainable 6-DOF joint. Each reference frame uses the conventions of anterior-posterior (AP), proximal-distal (PD), and medial-lateral (ML) for the X, Y, and Z axes, respectively. KR is the knee reference frame location and orientation in the tibia body. The socket reference frame (SR) is defined in the socket body. FR is the foot attachment reference frame in the socket body, and FRR is the foot flexion reference frame used to define the location of a single DOF joint for lumped foot flexion motion. C, Socket-limb joint configurations are listed, showing free coordinates for the accompanying configuration and description. The rigid model allows no motion at the socket-limb joint, the flex model allows flexion/extension only, the Pist model allows distal/proximal translation, the flex/Pist model combines the flex and Pist models, the 4-DOF model allows distal/lateral translation and rotation about all 3 axes, and the 6-DOF model is completely unconstrained

accounting for the relative displacements of the tracking markers on the pylon and prosthesis foot during motion capture. A more complex kinematic model of the prosthetic foot[7] could always be incorporated if desired.

2.2 | Experimental procedures

The socket-residuum model was evaluated for 3 test subjects. Motion data were collected during over-ground walking using an 11-camera optical motion capture system (Qualisys, Inc., Gothenberg, Sweden) to track reflective infrared markers attached to the test subjects at 240 Hz. Ground reaction forces were recorded at 2400 Hz using 3 flush-mounted strain gauge force platforms (OR6-5, AMTI, Inc. Watertown, MA, USA). Kinematic and ground reaction force data were both low-pass filtered (6 Hz) at the same cutoff frequency[18] using a dual-pass Butterworth digital filter.[19] The test subjects were all healthy individuals with left, unilateral transtibial amputations (Table 1).

Markers were placed on the test subjects by the same researcher across subjects for model scaling and motion tracking. Scaling markers included left and right acromion process, iliac crest, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, lateral and medial femoral condyles, lateral and medial malleoli, first metatarsal head, fifth metatarsal head, and the tip of the second toe. Markers on the left prosthesis were matched as closely as possible with corresponding markers on the intact side. Tracking markers included acromion processes, iliac crests, anterior

TABLE 1 Subject data

Subject Metrics	Subject 1	Subject 2	Subject 3
Height (m)	1.88	1.80	1.83
Mass (kg)	104.3	68.0	73.2
Gender	Male	Female	Male
Amputation	Left transtibial	Left transtibial	Left transtibial
Activity level	K-4	K-4	K-4
Prosthesis	Ossur variflex	Ossur elation	Ability dynamics rush 87
Residual/intact limb ratio	0.44	0.39	0.72
Preferred walking speed (m/s)	1.26	1.46	1.21
Cause of amputation	Trauma	Sarcoma	Congenital
Suspension type	Pin lock	Pin lock	Suction liner
Socket load bearing	Patellar tendon	Patellar tendon	End bearing

superior iliac spines, posterior superior iliac spines, toes, clusters of 4 markers on the thighs, shank (intact side), socket (affected side), and clusters of 3 markers on the heels of the shoes.

Marker data were first collected with the subjects standing in a static pose for model scaling purposes. Marker trajectory and ground reaction force data were then collected for level over-ground walking at the subjects' preferred speed (Table 1). In the IK solution process, more weight was placed on markers that are less susceptible to soft-tissue artifact, such as on the prosthesis/socket and the rigid clusters, while certain other markers, such as the markers over the greater trochanters, were used for scaling but were not used in IK. A successful experiment trial was defined as one where the subject stepped fully onto the 3 force platforms in a sequential pattern with alternating feet, but without looking down to target the platforms. Walking speed was measured with photogates placed 6 m apart. Three successful trials were recorded per subject so that within-subject ensemble averages could be calculated.

2.3 | Model preparation

Prior to scaling the generic model with the OpenSim scaling tool, the residuum in the model needed to be pre-scaled to properly reflect the specific level of amputation, accounting for the lack of scaling markers on the residuum. Using physical measurements of the subject, the residuum (lateral femoral epicondyle to the distal end of the transected limb) was calculated as a ratio of the length of the intact limb (lateral femoral epicondyle to lateral malleolus) (Table 1). Residuum and prosthesis inertial and mass properties were scaled in the model to reflect the residuum-to-intact limb length ratio.

After pre-scaling the residuum, the models were scaled using the OpenSim scale tool. Normally, each body would be scaled to a ratio of the distance between experimental markers to the distance between virtual model markers placed primarily at joint centers of rotation. Because experimental markers could not be mounted to the residual limb and affected knee center of rotation, the affected femur, residual tibia/fibula, and prosthesis bodies were scaled proportionally to the intact limb.

After scaling, model marker placement and socket joint reference frame position and orientation were optimized for the model for each subject using a recursive IK, marker placement routine. Anchor markers were first manually placed on the model sternum, heel center, and toe. Anchor markers are model tracking markers that are constrained in X, Y, and Z local coordinates during the marker placement optimization, except for the toe local X (anterior-posterior) coordinate, which was allowed to be optimized to account for inconsistencies in the placement of this marker on the shoe. The use of constrained anchor markers ensured that there was a unique solution to the marker placement optimization. The routine then modified all other tracking marker placements with 1-mm increments in the local X, Y, and Z coordinates sequentially, performing an IK analysis of a full gait cycle between each coordinate alteration, with the objective of minimizing marker tracking error during walking trials. The static position and orientation of the socket reference frame at the joint center were also optimized in the same routine to account for prosthesis alignment. This routine ensured a consistent approach for marker placement and prosthesis/socket alignment across all subject models.

2.4 | Data processing

IK and ID analyses were performed for the 6 residuum-socket joint DOF configurations (Figure 1C) on each of the 3 subjects. The *Rigid* joint configuration constrained all 6 DOFs. With respect to the socket reference frame shown in Figure 1B, the *Flex* configuration constrained all the DOFs except socket flexion/extension (rotation about the medial-lateral axis), while the *Pist* configuration constrained all DOFs except socket pistoning (translation along the proximal-distal axis). The *Flex/Pist* configuration combined free coordinates of *Flex* and *Pist,* and the *4-DOF* configuration combined the free coordinates of *Flex/Pist* with the added coordinates of adduction/abduction and residuum-socket rotation (rotations about the anterior-posterior and proximal-distal axes, respectively). Pistoning and flexion/extension were considered to be the most clinically relevant displacement; however, we included the 2 additional rotational coordinates as we have observed that some subjects exhibit visible angular displacements about 1 or both of these axes. Lastly, a *6-DOF* residuum-socket joint configuration was included to compare the unconstrained case with the other configurations and to examine the convergence of model marker tracking error with increased model complexity. With the exception of the *6-DOF* configuration, anterior-posterior and medial-lateral translational movements in the socket reference frames were constrained because the residuum segment would be kinematically under-constrained otherwise, and further, these were considered to be the least clinically relevant motions. Even though these motions were constrained, it would still be possible to compute the associated forces if interested.

Average IK marker tracking root mean square (RMS) and maximum tracking error values were calculated across all trials for all residuum-socket joint DOF configurations. Marker error results were presented in absolute units and normalized to the *Rigid* model configuration to show percent error reduction in other model configurations. IK and ID data were normalized to percent stride for each trial. Averages and standard deviations were then calculated across trials for each joint configuration. Model performance was evaluated in terms of trade-offs between marker tracking error and the resulting joint kinematics and kinetics.

3 | RESULTS

Model marker tracking RMS error generally decreased with increasing model complexity (Figure 2). On average, the *4-DOF* joint configuration had a 45% reduction in marker RMS error as compared with the *Rigid* configuration. However, there was almost no difference in the marker RMS error for the unconstrained *6-DOF* joint configuration relative to the *4-DOF* condition (Figure 2). Not only did the *6-DOF* condition not yield any meaningful reduction in marker tracking error over the *4-DOF* case, but the *6-DOF* model resulted in unrealistic residuum-socket kinematics. This is a result of the 6-DOF model being under-constrained during the IK process due to a lack of tracking markers on the residual limb or kinematic constraints within the joint and could produce an infinite number of kinematic solutions to the global least squares optimization. The *4-DOF* model was selected to provide the best compromise for representing the residuum-socket interface, based on low marker tracking errors and results for the socket joint and anatomical joints that agree with data from the literature (see Results and Discussion sections). Therefore, subsequent kinematic and kinetic results are presented based on the *4-DOF* model.

Average joint angles (Figure 3) and moments (Figure 4) for walking at preferred speed are presented for the affected limb with energy storage and return (ESR) prosthesis and contralateral intact limb for the *4-DOF* model (dark and light solid lines, respectively). Complementary *Rigid* model results for the affected and intact limbs are included to highlight the differences in calculated movements and forces: compare dashed to solid dark lines (ESR) and dashed to solid light lines (Intact). Comparing the *Rigid* and *4-DOF* model results, offsets can be seen throughout the gait cycle, most notably in the affected limb. The calculated kinematics for the prosthetic foot differed maximally by approximately 2°, whereas the affected knee and affected hip each differed maximally by approximately 9° (Figure 3). Calculated kinetics differed maximally by approximately 0.1 N-m/kg in the affected limb (Figure 4). These maximal differences occur primarily at the peak values, occurring at maximum knee flexion during swing and around 50% gait for the hip moment.

Figure 5A to L presents the *4-DOF* residuum-socket joint kinematic results, comparing across subjects. Pistoning (Figure 5A–C) shows compression of the residuum inside of the socket immediately after heel strike. The amount of pistoning varies considerably between subjects, with maximum compression displacement of 25 mm during the stance phase for subject 1, and maximum elongation displacement of 33 mm during the swing phase for subject 3. Flexion/extension (Figure 5D–F) revealed similar inflection point patterns with different ranges between subjects; however,

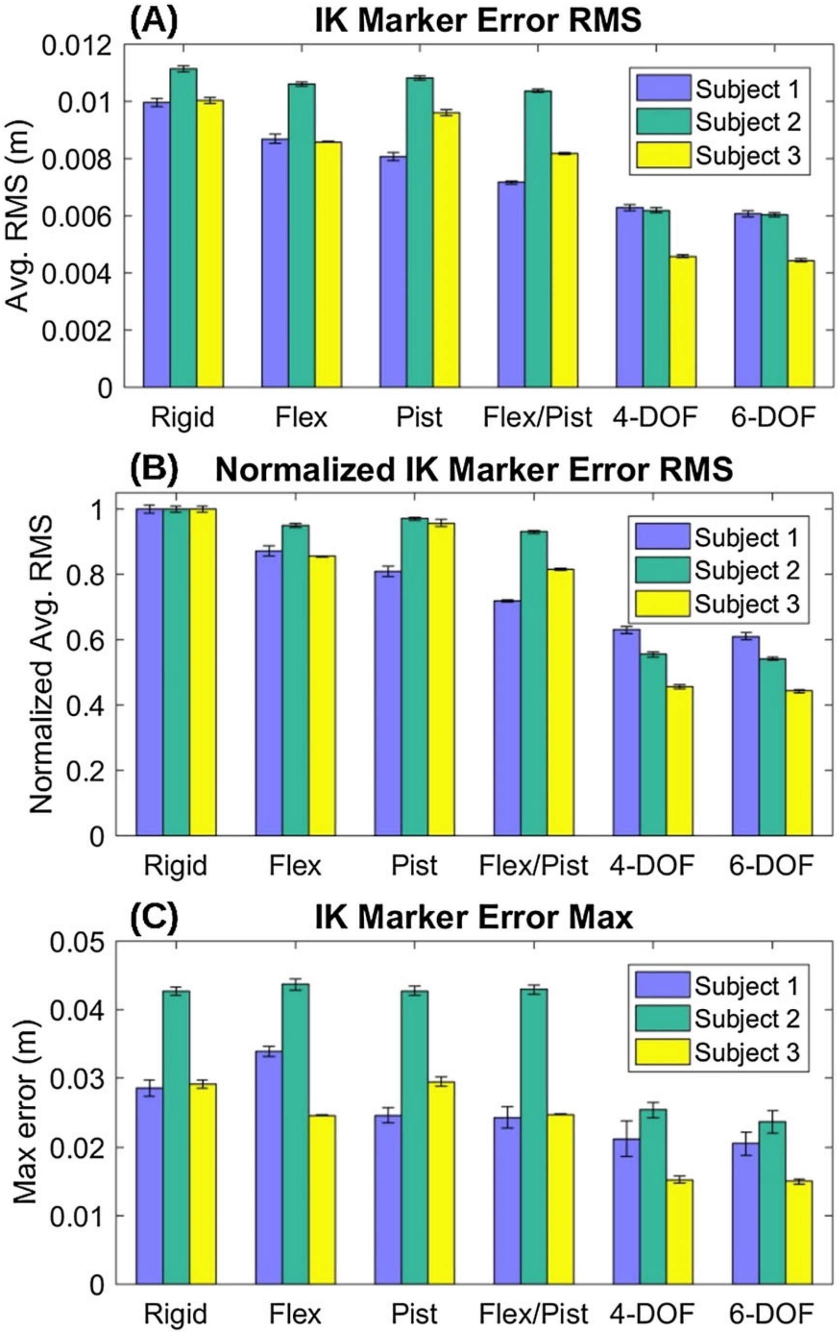


FIGURE 2 Model marker RMS errors from inverse kinematics are shown averaged across preferred speed walking trials for each model variation tested. The data in A, are average values, and the data in B, are normalized to the rigid model marker RMS error for each subject individually, which exhibited the largest error value of all models. The data in C, reflect the maximum tracking error for a single marker averaged between trials. RMS error values averaged between subjects ranged from 0.57 cm (4-DOF) to 1.04 cm (rigid). The fully constrained 4-DOF socket-limb joint reduced marker error by 45% on average. The difference in marker error between the 4-DOF joint and the under-constrained 6-DOF joint typically fell within between-trial uncertainty

subject 2 had notable differences in peak magnitudes. Abduction/adduction (Figure 5J–L) had the smallest rotational range of motion (minimally 2 to maximally 5° between subjects), while axial rotation (Figure 5G–I) had the most variable range of motion (minimally 16 to maximally 38°).

4-DOF residuum-socket joint kinetic results are shown in Figure 6, corresponding to the kinematic data presented in Figure 5. Pistoning forces (Figure 6A–C) exhibit a pattern with 2 peaks during the stance phase with a trough in between. There was an extension moment about the medial-lateral axis for approximately the first 20% of the gait cycle, and then a flexion moment for the rest of the stance phase, with a moment magnitude close to zero for the last 40% of the gait cycle, corresponding to the swing phase (Figure 6D–F). The flexion-extension moments were similar in pattern to the ankle joint moments (Figure 4A–C), but the peak magnitudes in mid-stance varied by up to 0.49 N-m/kg. The axial rotation moment averages (Figure 6G–I) showed consistent patterns throughout the stride; however, the standard deviation was noticeably larger during late stance. Abduction/adduction moments (Figure 5J–L) present a similar pattern across subjects.

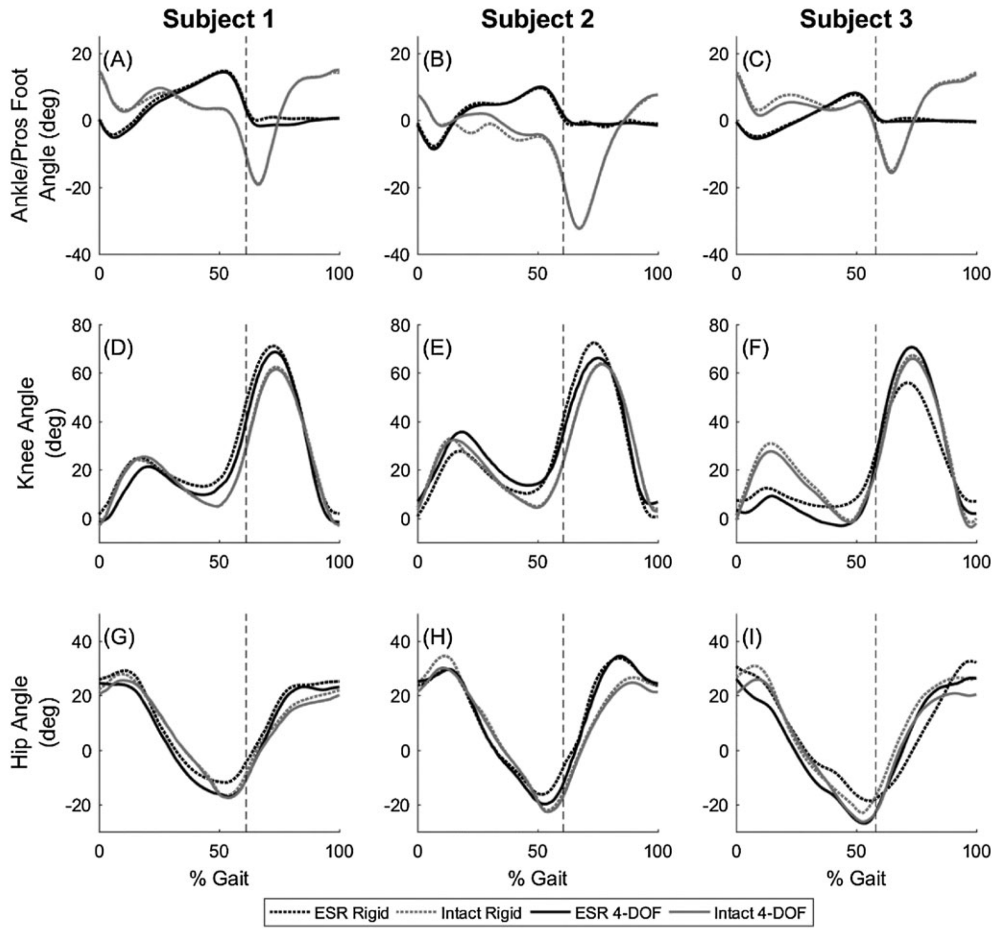


FIGURE 3 The resultant average kinematic data are shown for the ankle/prosthetic foot A–C, knees D–F, and hips G–I, for each subject. In each plot, the dark solid lines are average data for the amputated limb with a 4-DOF socket model and energy storage/return (ESR) foot, the light solid lines are average data for the intact limb, the dark dashed lines are average data for the amputated limb calculated with a rigid socket model, and the light dashed lines are average data for the contralateral intact joint. End of stance is marked based on the prosthesis side limb

4 | DISCUSSION

In this study, a model of the residuum-socket interface was included as part of 3-dimensional gait analysis for persons with transtibial amputation. The residuum-socket joint was modeled as having 1 translational and 3 rotational DOFs, which is more biomechanically relevant than the current practice of modeling this interface as a rigid joint, either implicitly or explicitly, or the alternative of using a completely unconstrained joint. A 4-DOF joint was found to provide the best compromise between faithfully representing the measured motion (ie, low marker tracking error), while yielding kinematic results with no violations of physical constraints, such as the residual limb penetrating through the socket. This approach should allow for more accurate assessment of whole body movement, while also providing insights on residuum-socket mechanics.

Examining marker tracking RMS error (Figure 2), both reduction and convergence are seen, indicating improved model fit with increased DOFs across all subjects up to the 4-DOF model, with almost no differences in marker tracking error for the 6-DOF model relative to the 4-DOF model. On average, the marker RMS error average was reduced by 45%. Peak marker errors were reduced, on average, from 3.3 to 1.9 cm, indicating that overall marker tracking was well within the acceptable range expected in a marker-based IK analysis.[20,21] The difference between marker RMS and peak errors for the 4-DOF and 6-DOF models was negligible. This suggests that socket motion is adequately accounted for with the 4-DOF joint. Also, it shows that the 4-DOF joint is not over-constrained, which could affect the accuracy of the calculated kinematics for the rest of the body if it were.

Observing the whole body kinematics and kinetics (Figures 3 and 4), these data show an asymmetric gait as is commonly seen in the gait of persons with unilateral amputation, but generally agree with data found in the literature.[6] Comparing the calculated residuum-socket kinematics and kinetics across subjects (Figures 5 and 6), it is seen that the residuum-socket joint is more variable from person-to-person than intact joints both in magnitude and overall

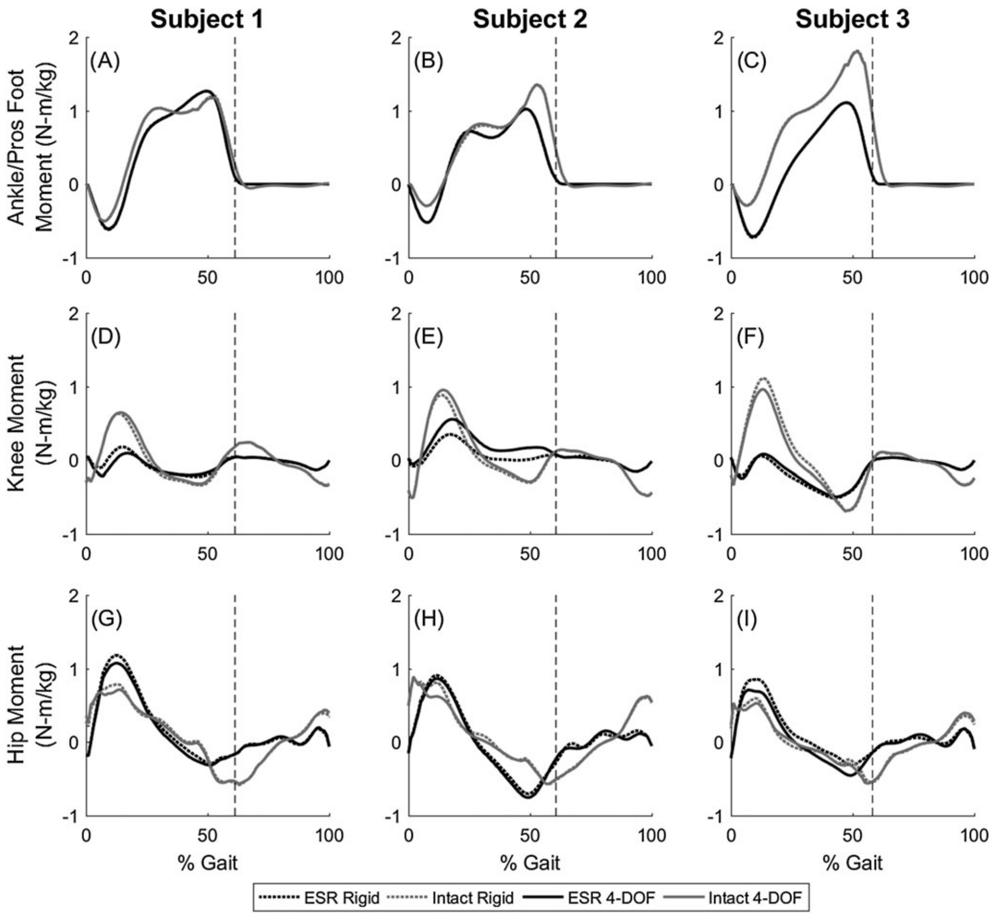


FIGURE 4 The resultant average kinetic data are shown for the ankle/prosthetic foot (A–C), knees (D–F), and hips (G–I)

pattern. This is most likely due to the fact that the interaction between the residuum and prosthesis is heavily influenced by the suspension type, amputation level, amount of soft tissue, quality of socket fit, perceived comfort, and other variables that impact one's performance while using a prosthesis.

Using the socket model, it can be seen that the pistoning motion compressed during stance and rebounded during swing as would be expected. Overall, the general pattern and range of pistoning motion agree with motions that have been found in previous studies.[1,2,4,10] The resulting pattern of motion for socket flexion/extension was generally consistent with results reported in the literature as well.[4,10] The large magnitude of axial rotation indicates that this coordinate may be a previously unappreciated aspect of gait mechanics in people with lower limb amputation. This substantial motion would not be captured in experimental or simulation analyses that do not explicitly account for motion at the residuum-socket interface. Lastly, abduction/adduction displacements have the smallest range of motion of the 4-DOFs. Comparing these motions between subjects shows they are highly variable and should only be taken as an estimate of socket motions based on the data gathered during routine gait analysis. Any anterior-posterior and medial-lateral translational motions are coupled into the other 4-DOFs which are a likely source of error and would increase with poorer fitting sockets. However, these calculated results can be interpreted as a fair estimate of the residuum-socket motion due to the small standard deviation seen across trials for each subject.

When examining the generalized forces calculated using ID, similar patterns are observed in all 3 subjects (Figure 6A–L). The pistoning force has 2 peaks with a trough in between, similar to the vertical ground reaction force, and shows a low standard deviation for each subject. Overall, the patterns and ranges of values of the socket kinetics agreed with existing data reported in other studies.[22,23] Axial rotation moment data (Figure 6G–I) were consistent across datasets. However, the higher standard deviation indicates that this coordinate exhibits considerable variability on a step-to-step basis. This variability may be a result of the axial rotation counter moments being generated from shear forces and large tissue deformations within the socket (ie, no structural boney hard stop), giving the subject little control of the counter moments needed for stability. Although the axial rotation moment magnitude is low during stance in comparison to other coordinates, axial rotation may play a key role in some locomotion strategies considering the large movement range (Figure 5G–I).

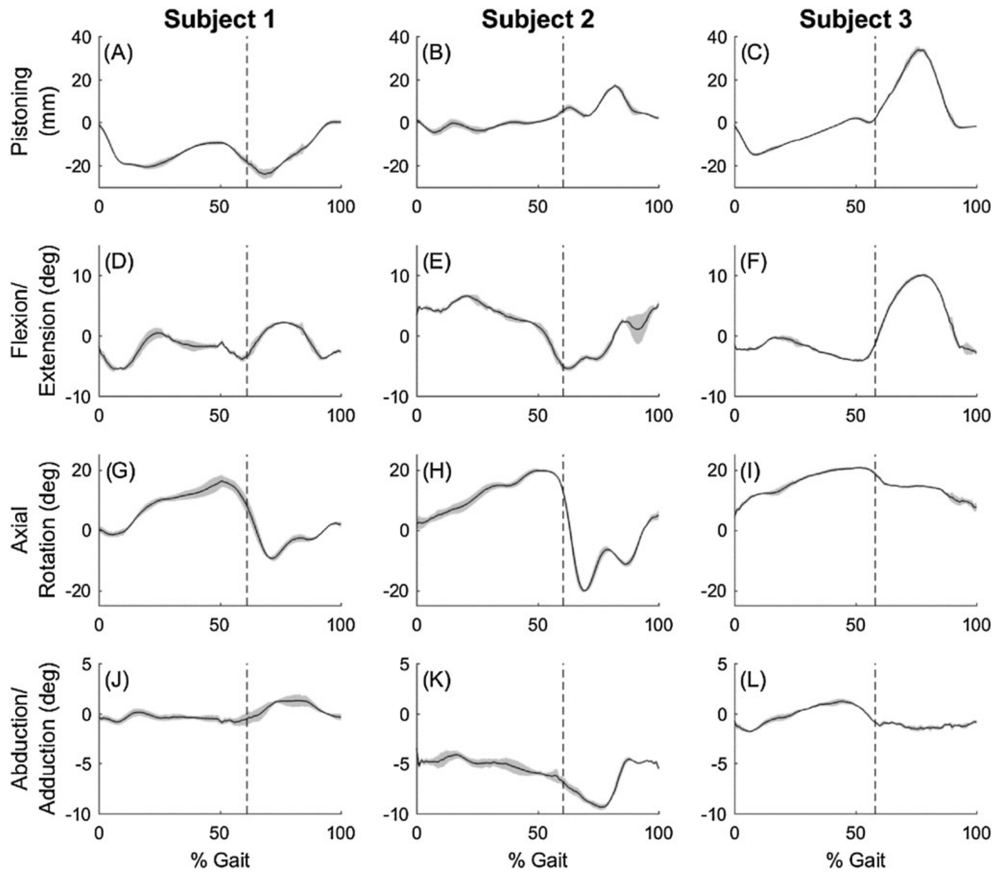


FIGURE 5 The resultant average kinematic data with 1 standard deviation are shown for the 4-DOF socket-limb model for each subject

Quantifying residuum-socket kinematics and kinetics using only non-invasive, passive marker-based motion capture data is a significant advantage of the presented method over other possible approaches. The methodology does not require additional imaging systems, such as ultrasound, magnetic resonance imaging, or radiological imaging.[4,5,9-11] Such imaging systems add complexity to test procedures and are cumbersome to incorporate into routine gait analyses. Also, the model-based approach presents almost no risk to subjects compared with other methods such as x-rays and video-fluoroscopy that use potentially dangerous radiation and can only be used in limited cases.[10,11]

A limitation of the presented method is that translational movements in the anterior-posterior and medial-lateral directions within the socket must be constrained statically for the IK analysis to converge to a unique solution, even though the residuum-socket joint has 6 DOFs in actuality. If the joint was modeled with 6 DOFs in this type of analysis, it would be unconstrained and possibly lead to non-physiological results for residuum-socket mechanics. An alternative approach would be to use soft tissue constraints (eg, Gasparutto et al[24] and Skipper Andersen et al[25] but this would depend on having good estimates of the stiffness and damping of the soft tissues in the residual limb, which are likely to be variable across subjects. Although the *4-DOF* joint model does provide a window into the residuum-socket interaction, results of the calculated free coordinate movements are influenced by the actual anterior-posterior and medial-lateral translational movements which are not captured. This simplification introduces a source of error, but the actual anterior-posterior and medial-lateral movements are likely to be small. Thus, the error is also likely to be small. To reduce the errors, a method is needed to directly measure bone motion by conducting a concurrent marker based and imaging study in the future. It should also be noted that while the translations in the anterior-posterior and medial-lateral directions cannot be quantified with our approach, the forces in these directions can still be calculated if desired.[26]

5 | CONCLUSION

This paper reports on the development and evaluation of an approach for directly incorporating the residuum-socket interface into gait analysis procedures for persons with limb loss walking with a transtibial prosthesis. Our approach uses

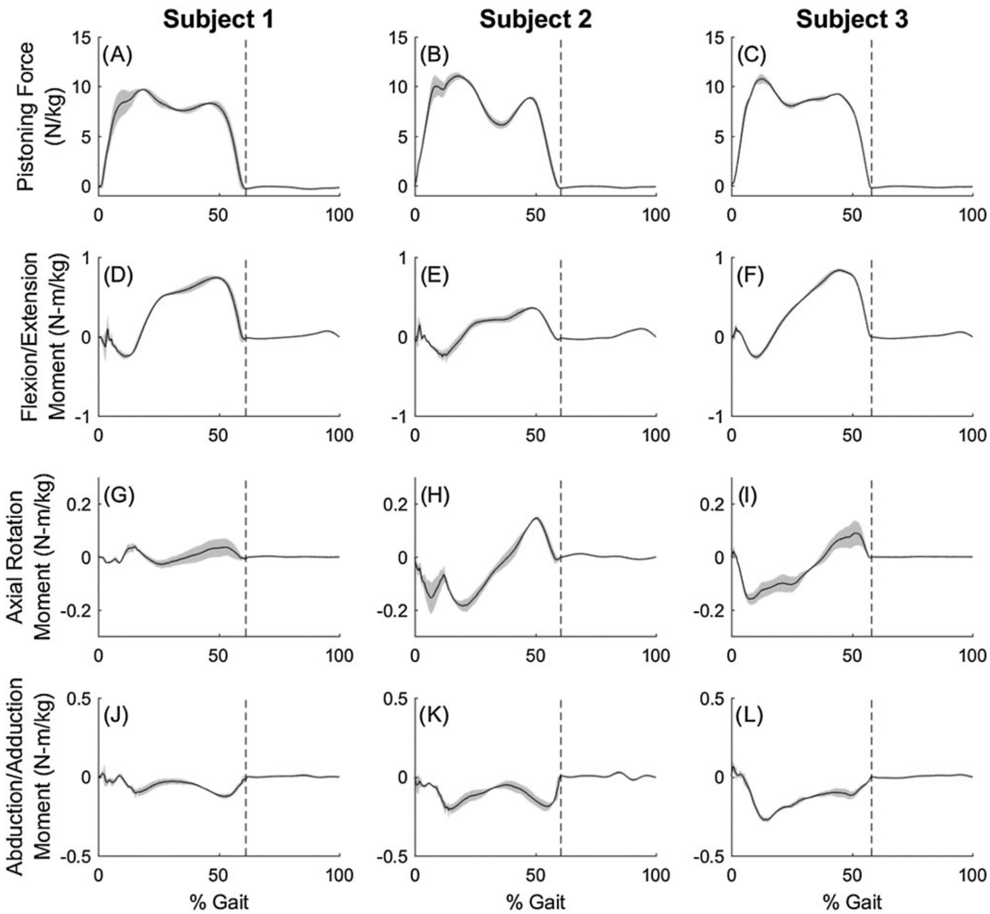


FIGURE 6 The resultant average kinetic data with 1 standard deviation are shown for the 4-DOF socket-limb model for each subject

a least-squares global optimization IK approach to estimate the motion in 4 of the 6 DOFs at the residuum-socket joint. Results for residuum-socket motion and loading were generally in agreement with measurements reported in the literature, and future efforts will focus on further validating the model predictions using imaging techniques. Accounting for the motion and loading at the interface between the prosthesis and the human user can lead to a better understanding of the relationship of socket fit to overall prosthetic performance and lead to improvements in prosthesis and socket design.

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