Asociación Argentina
de Mecánica Computacional

Mecánica Computacional Vol XXXVII, págs. 1511-1520 (artículo completo)
A. Cardona, L. Garelli, J.M. Gimenez, P.A. Kler, S. Márquez Damián, M.A. Storti (Eds.)
Santa Fe, 5-7 Noviembre 2019

ON THE IMPROVEMENT OF THE SCALING OF THE TRANS-TIBIAL AMPUTEE MUSCULOSKELETAL MODEL BY USING A VIRTUAL MARKER APPROACH

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Keywords: Musculoskeletal model, Scaling, Amputee gait.

Abstract. Modern marker-based motion capture techniques applied in clinical gait laboratories rely on the conventional gait analysis model as a standard tool for studying normal and pathological gait. It uses a Direct Kinematics (DK) method to calculate joint kinematics. However, current techniques are limited when used to analyse the biomechanics of amputee gait. This is because the residuum-socket interface is usually assumed to be rigid despite the considerable motion of the residual limb inside the socket which is known to occur. This relative motion is not captured in routine gait analysis due to the inability to place tracking markers inside the socket. In contrast to conventional models based on DK, musculoskeletal modelling software solve a global-optimisation for 'pose estimation' using an Inverse Kinematics (IK) method. Thus, while the residuum-socket interface should be treated explicitly as another joint in the system, this presents challenges when using traditional marker-based kinematic techniques. Previous studies of biomechanics in amputee gait have used various sensing technologies to directly record residuum-socket motion. However, such methods are complex and require costly or highly specialised equipment, these methods are therefore impractical for routine gait analyses (LaPrè et.al., *Int J Numer Method Biomed Eng*, 34(4):29–36 (2018)). To address this limitation, we develop a whole-body musculoskeletal model including a residuum-socket joint with a least-squares global-optimisation method to study the kinematics of this interface, simultaneously with traditional joint angles during amputee walking. Experimental walking data from 4 subjects with unilateral trans-tibial amputations were used to demonstrate the utility of the virtual marker approach to scale a generic amputee musculoskeletal model during walking. The effects of conventional surface marker locations and the virtual marker approach to scale this generic musculoskeletal model were examined, and the resulting residuum-socket generalised motions for each model configuration were compared for all subjects.

1 INTRODUCTION

Computational models may become a tool that help to improve the clinical gait analysis and treatment of gait abnormalities, providing more effective strategies for therapeutic management (Ravera et al., 2014). An area of clinical care that can benefit from this approach is orthopaedics. Amputee gait, one of the more common gait pathologies with relatively high prevalence rates in many countries. More amputations are performed each year for people with vascular disease, secondary to trauma and bone or joint malignancy which leads to the increment of annual acute and post-acute medical care costs associated with caring for amputees in the health-care policies of each country (Sagawa et al., 2011). Musculoskeletal models are largely used for studying the muscle behaviour during walking in unimpaired subjects (Erdemir et al., 2007). However, even though hundreds of clinical gait laboratories around the word have used classical gait analysis (i.e. conventional gait analysis modelling proposed by Kadaba et al. (1990); Davis et al. (1991)) as gold standard for the diagnosis of patients, musculoskeletal models are not yet applied in clinical practices with the same frequency (Hicks et al., 2015). Amputee gait analysis is most frequently analysed through conventional gait analysis modelling (Silverman and Neptune, 2012; Fey et al., 2013; LaPre et al., 2014). However, applying a conventional model to prosthetic gait implies powerful assumptions. In particular, the conventional model uses a Direct Kinematics (DK) method to calculate joint kinematics during walking while, musculoskeletal models perform a global-optimisation for 'pose estimation' using an Inverse Kinematics (IK) method which has been shown to reduce soft tissue artefacts (Kainz H. et al., 2017).

In addition to the limitations of DK, conventional gait models are limited when used to analyse the biomechanics of amputee gait. This is because the residuum-socket interface is usually assumed to be rigid despite the well-known and considerable motion of the residual limb inside the socket. Moreover, this relative motion is not captured in routine gait analysis due to the inability to place tracking markers inside the socket (LaPrè et al., 2018). Thus, while the residuum-socket interface should be treated explicitly as another joint in the system, this presents challenges when using traditional marker-based kinematic techniques.

Previous studies of biomechanics in amputee gait have used various sensing technologies to directly record residuum-socket motion (McGrath et al., 2017). However, such methods are complex and require costly or highly specialised equipment, these methods are therefore impractical for routine gait analyses. Recently, LaPrè et al. (2018) have developed an open-sourced musculoskeletal model available in OpenSim (https://simtk.org/projects/bkamputee_model) to estimate the motion and loading at the residuum-socket interface as part of standard gait analysis procedures. However, this model requires a non-automatic scaling process that could impede their use in clinical gait analysis due to the time and prior knowledge in musculoskeletal modelling required to scale the model.

The aim of this study was to develop a whole-body musculoskeletal model including a residuum-socket joint with a least-squares global-optimisation method to study the kinematics of this interface, simultaneously with traditional joint angles computations in trans-tibial amputee gait. In particular, a virtual marker approach to scale a generic amputee musculoskeletal model was proposed to automate the model scale process and thus increase the use of this models in clinical settings.

2 EXPERIMENTAL PROCEDURE

2.1 Participants and data collection

Four participants that walk at self-selected comfortable walking speed were considered in this study. The subjects had left, unilateral trans-tibial amputations without concurrent pathologies (Table 1). A successful trial was defined as one where the subject stepped fully onto the recording runway in a sequential pattern with alternating feet, but without looking down to target the runway. Six successful trials were recorded per subject so that within-subject ensemble averages could be calculated (Kadaba et al., 1989). A total of twenty four (24) gait trials were analysed.

Subject Metrics	Subject 1	Subject 2	Subject 3	Subject 4
Height (m)	1.785	1.733	1.761	1.753
Left leg length (m)	0.936	0.935	0.955	0.920
Right leg length (m)	0.914	0.920	0.950	0.915
Mass (Kg)	73.00	57.00	71.00	86.50
Gender	Female	Female	Male	Male
Amputation	Left trans-tibial	Left trans-tibial	Left trans-tibial	Left trans-tibial
Activity level	K-4	K-4	K-4	K-4
Prosthesis[a]	Hydraulic ankle	Hydraulic ankle	Hydraulic ankle	Hydraulic ankle
Suspension type	Pin lock	Suction liner	Vacuum	Suction liner
Stump length (m)	0.540	0.680	0.630	0.630
Thigh length of residual limb (m)	0.520	0.480	0.540	0.520

Table 1: Description of participants.

[a] All participants walk with Echelon prosthesis from Blatchford Groups, United Kingdom (https://www.blatchford.co.uk/endolite/echelon/).

Motion data were collected during over-ground walking using a 12-camera optical motion capture system running at $240\ Hz$ (Qualisys, Inc., Gothenberg, Sweden) to track reflective infrared markers attached to the test subjects. Reflective markers were placed on each participant according to a modified Helen Hayes marker set (Tabakin, 2000). Markers on the prosthetic site were matched as closely as possible with corresponding markers on the intact side. Also, marker data with the subjects standing in a static pose were collected for model scaling purposes. Marker tracked data was low-pass filtered at $6\ Hz$ cutoff frequency using a dual-pass Butterworth digital filter.

2.2 Musculoskeletal modelling

The prostheses-specific musculoskeletal model was developed by modifying the *gait2392* model, a standard gait model provided in OpenSim 4.0 (Delp et al., 2007). A residual tibia/fibula (stump) segment was created, and it was connected to the specific prosthesis through a custom residuum-socket joint. The musculoskeletal geometry was based on Delp (1990), while the inertial properties of the residual leg shank segment were adjusted based on the work of LaPrè et al. (2018). In order to model the prosthesis, a specific model was selected and its technical characteristics were used to model the segment.

The prostheses-specific musculoskeletal model has 24 degrees of freedom (Dof) and 82 musculotendon actuators to represent 66 muscles in the lower extremities and torso (Figure 1). The Dof in this model included three translations and three rotations of the pelvis; three ball and socket joints, one located at the third lumbar vertebrae to model the interaction between pelvis and trunk and two at each hip joints; a custom joint with coupled translations and rotations at each knee; and a revolute joint at the intact ankle and the prosthetic ankle. The residuum-socket joint was modelled as a custom joint in accordance with LaPrè et al. (2018). Residuum-socket joint included 1 translation and 3 rotational Dof. This configuration was selected because it provided the best compromise between faithfully representing the measured motion (i.e., low marker tracking error) with no violations of physical modelling constraints and anatomical joints that agree with data from the literature (LaPrè et al., 2018).

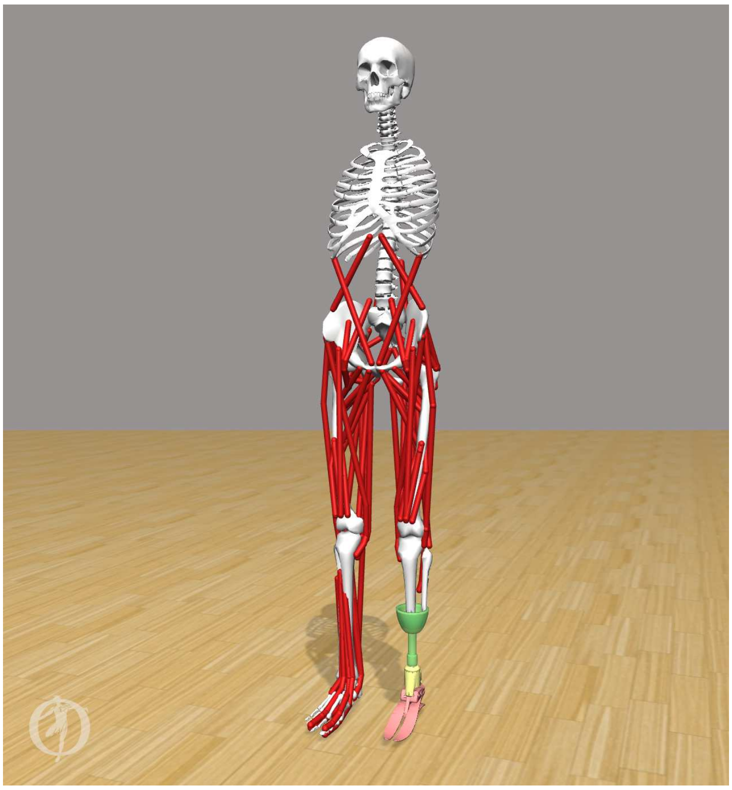

Figure 1: Snapshot of the prostheses-specific musculoskeletal model developed.

2.2.1 Residuum-socket's virtual marker approach

In the computational modelling context applied to gait analysis, a virtual marker is defined as a three-dimensional point which seeks to represent the position of a marker where it has not been actually placed in the static pose. A virtual marker was defined in order to represent the residuum-socket joint in standing (Figure 2). This virtual marker was created as a traslation along the long axis of the shank's local coordinate system equal to the Residual limb length (the difference between stump length and the thigh length of residual limb) from the knee joint position in the static pose following the equation (1).

$$\text{Residuum-socket} = (\text{Knee joint}) - (\text{Residual limb length}) \cdot \vec{k}_{shank} \,. \qquad (1)$$

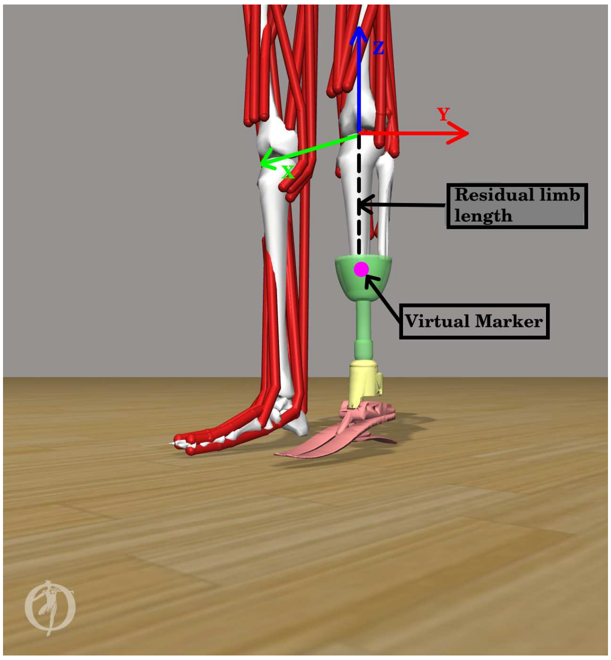

Figure 2: Schematic representation of the residuum-socket virtual marker.

2.3 Data processing

The prostheses-specific musculoskeletal model was scaled using the OpenSim scale tool. The scaling process is performed based on measured distances between marker positions obtained from standing data and the residuum-socket virtual marker defined in section 2.2.1. The dimensions of each segment in the musculoskeletal model are scaled so that the distances between its markers match the distances between the experimental markers recorded by the motion capture system on static pose (Kainz H. et al., 2017).

Inverse kinematics (IK), from the experimental marker trajectories, were used to calculate joint angles. In the IK solution process, more weight was placed on markers that are less susceptible to soft-tissue artefacts, such as on the pelvis and feet markers, while residuum-socket virtual marker was used for scaling but were not used in IK. The analyses were performed for the 6 trial on each of the 4 participants. Model performance was evaluated in terms of the coefficient of multiple correlation to examine the intrasubject repeatability on the joint results (Kadaba et al., 1989). In accordance with the results of previous studies (Sartori et al., 2013; Ravera et al., 2019) this metric was considered acceptable if $0.7 \leq CMC \leq 1.0$. Average IK marker tracking root mean square (RMS) and maximum tracking error values were calculated across all trials of each participant.

3 RESULTS AND DISCUSSION

Table 2 presents the performance of each joint included in the subject-specific amputee musculoskeletal model. All model's Dof have CMC above 0.74 which shows that the use of the virtual marker approach lead to acceptable intrasubject repeatability on the subject-specific amputee musculoskeletal modelling.

	Dof	Subject 1	Subject 2	Subject 3	Subject 4
CMC intact limb	Hip Flex/Ext	0.9850	0.9447	0.9945	0.9957
	Knee Flex/Ext	0.9792	0.8612	0.9921	0.9920
	Ankle Flex/Ext	0.9500	0.7714	0.9545	0.9898
CMC prosthetic limb	Hip Flex/Ext	0.9968	0.9818	0.9875	0.9942
	Knee Flex/Ext	0.9967	0.9524	0.9894	0.9868
	Pros ankle Flex/Ext	0.9961	0.9422	0.9645	0.9910
	Socket Flex/Ext	0.9799	0.9036	0.8242	0.8230
	Socket Add/Abd	0.9744	0.9226	0.9665	0.9164
	Socket Ext/Int	0.7427	0.8680	0.9053	0.9473
	Socket trans	0.9914	0.9241	0.9907	0.9775

Table 2: Coefficient of multiple correlation (CMC) of each joint of the prostheses-specific musculoskeletal model. The performance of Flexion-Extension of hip, knee and ankle joints for both intact and amputee limb and all degrees of freedom of the socket joint are presented.

Average of marker tracking RMS and maximum errors (Figure 3) are presented for each subject walking at self-selected speed. In agreement with the results of LaPrè et al. (2018), the amputee musculoskeletal model presents the marker tracking RMS and the marker tracking maximum errors less than $6.5\ mm$ and $1.75\ cm$, respectively. In additions, these error values indicate that overall marker tracking was well within the acceptable range in an IK analysis (https://simtk-confluence.stanford.edu:8443/display/OpenSim/Checklist+-+Evaluating+your+Simulation).

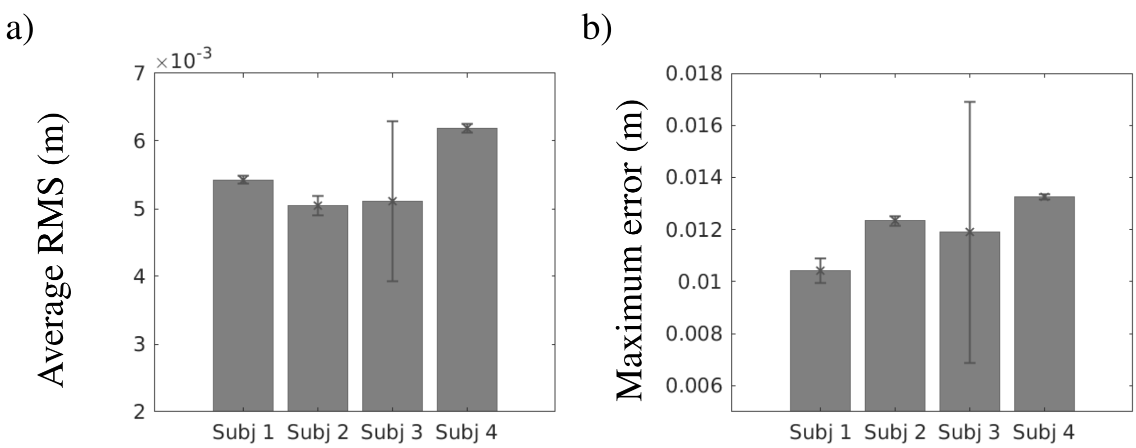

Figure 3: Mean and standard deviation of a) marker tracking RMS error and, b) marker tracking maximum error during walking for all musculoskeletal simulations of each subject.

Figure 3 shows that amputee subjects that use pin lock and suction lines as suspension for their prosthesis (see Table 1) have better simulation results with smaller variance of marker tracking errors in contrast with vacuum suspension.

Mean and standard deviation of joint angles for walking of amputee subjects are presented for intact and prosthetic limbs in Figure 4.

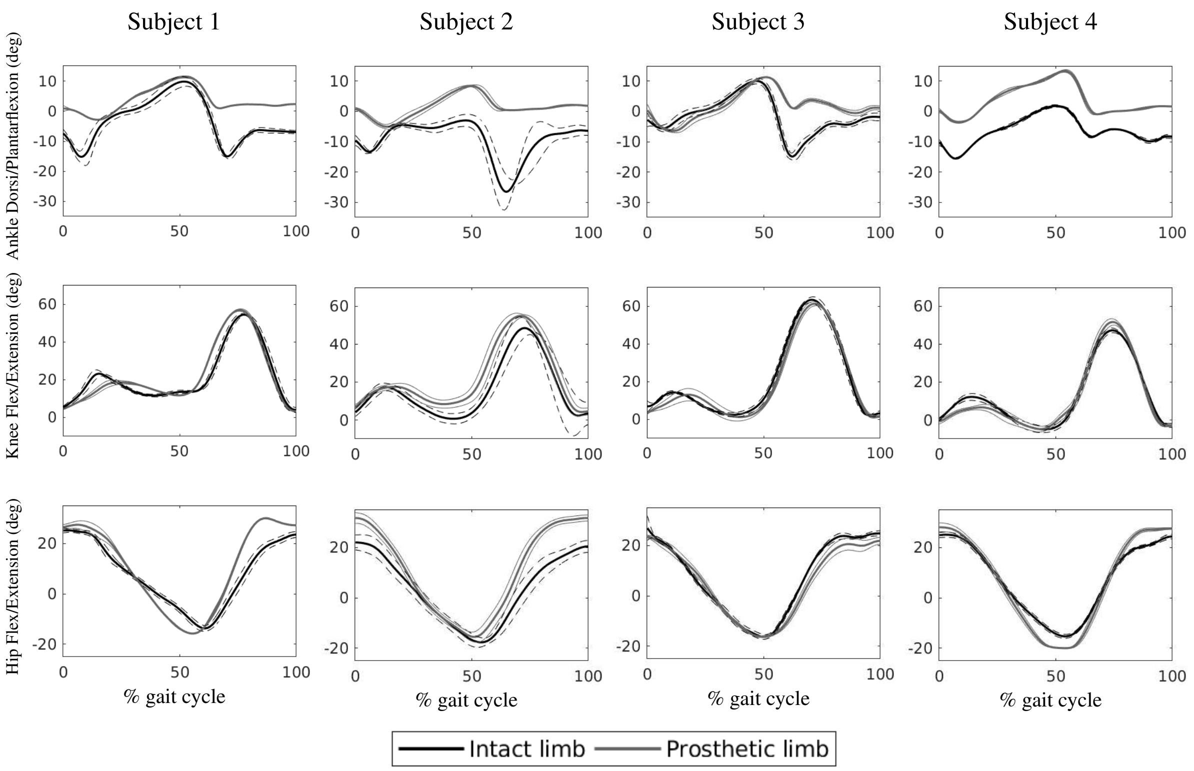

Figure 4: Mean (solid lines) and standard deviation (dash lines) of hip, knee, ankle and prosthetic ankle flexion-extension throughout the gait cycle.

Finally, residuum-socket joint kinematic results are shown in Figure 5. Immediately after heel strike, the residuum presents a negative translation (compression) over the socket around 2 cm. Amputee subjects with pin lock (subjects 1) and suction liner (2 and 4) have different translation pattern when compared to pin lock (subject 3) which compressed during stance and rebounded during swing. In particular, vacuum suspension present smaller vertical displacement of socket but suction liner presents a compression pattern that follows the body weight transition during walking and it could create better suspension, fit, and gait performance of amputee subjects (Gholizadeh et al., 2014). Also, the general pattern and range of socket translation motion agree with motions that have been found in previous studies (LaPrè et al., 2018).

Also, Figure 5 shows that the socket flexion/extension of subjects 1, 2 and 4 were between -2^o and -12^o while subject 3 have a socket flexion/extension between 0^o and 5^o. Socket abduction/adduction had the smallest rotational range of motion, while socket internal/external rotation had the biggest range of motion.

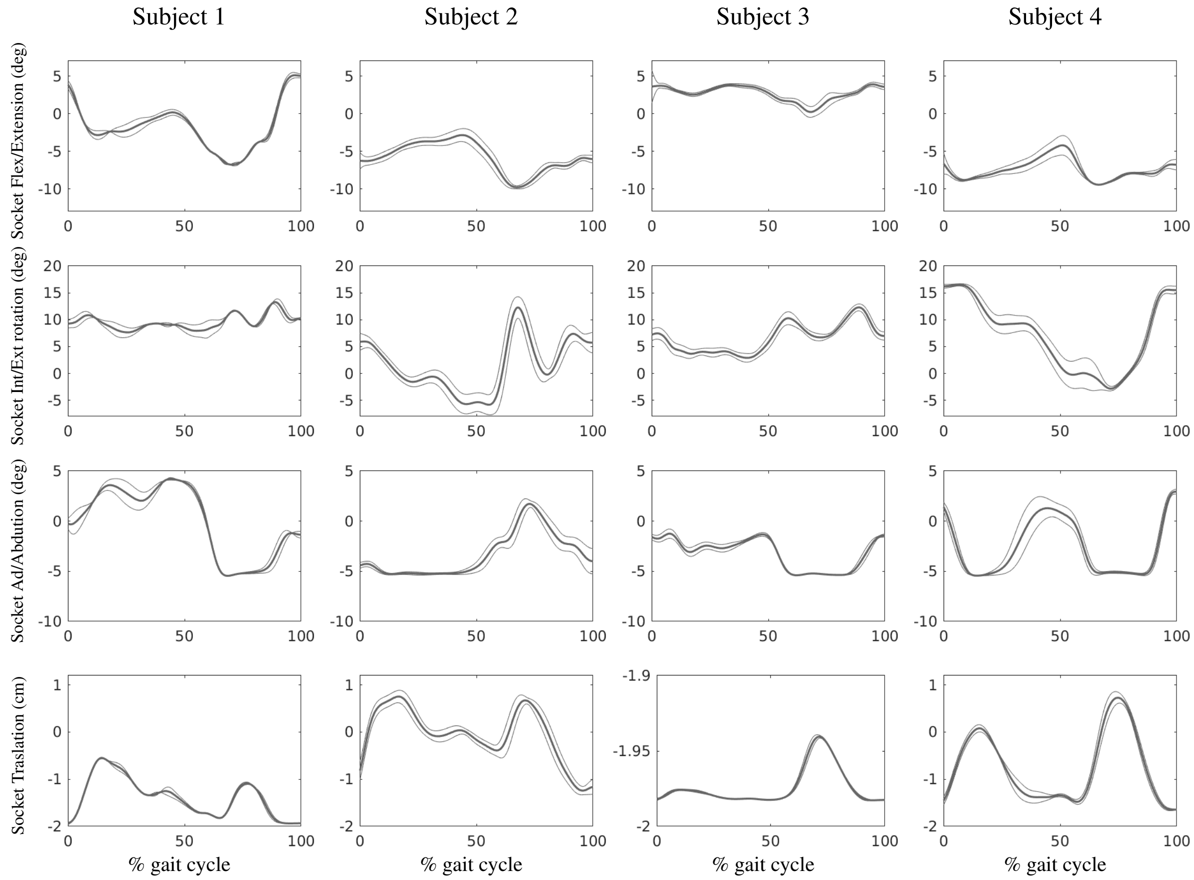

Figure 5: Mean (solid lines) and standard deviation (dash lines) of the residuum-socket joint throughout the gait cycle for each subject.

4 CONCLUSIONS

The use of the virtual marker approach to represent the residuum-socket joint in the scaling process of subject-specific amputee musculoskeletal model is a significant advantage of the presented model over similar models that use manual scaling processes. Also, this approach was shown to be a robust and automatic process with acceptable performance; it had the highest intrasubject repeatability in the kinematic results of all degrees of freedom included in the musculoskeletal model. This result is of particular importance when the model is used in clinical gait laboratories. Moreover, the virtual marker approach presents almost no risk to subjects compared with other methods such as x-rays and video-fluoroscopy that use potentially dangerous radiation and cannot be used in a typical clinical setting. Future work will include larger number of subjects to generalise the results.

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