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► **To cite this version:**

Paul-André Deleu, Alexandre Naaim, Thibaut Leemrijse, Raphaël Dumas, Bernhard Devos Bevernage, et al.. Impact of foot modeling on the quantification of the effect of total ankle replacement: A pilot study. *Gait & Posture*, 2021, 84, pp.308-314. <10.1016/j.gaitpost.2020.12.027>. <hal-03108323>

HAL Id: hal-03108323

<https://hal.science/hal-03108323v1>

Submitted on 13 Jan 2021

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HAL Authorization

Journal Pre-proof

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PII: S0966-6362(20)30694-9
DOI: <https://doi.org/10.1016/j.gaitpost.2020.12.027>
Reference: GAIPOS 8017
To appear in: *Gait & Posture*
Received Date: 15 July 2020
Revised Date: 17 November 2020
Accepted Date: 22 December 2020

Please cite this article as: { doi: <https://doi.org/>

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**Impact of foot modeling on the quantification of the effect of total ankle
replacement: a pilot study**

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Word count : 3,120 words

Abstract : 278 words

HIGHLIGHTS

- ‘Foot model’ has a significant effect on range of motion and power estimates;
- One-segment foot models overestimate ankle joint mechanics in pathological gait;
- Effect of total ankle replacement on ankle kinematics is limited;
- Total ankle replacement gait studies may benefit from multi-segment foot modelling.

ABSTRACT

Background: Kinematic and kinetic foot models showed that computing ankle joint angles, moments and power with a one-segment foot modeling approach alters kinematics and tends to overestimate ankle joint power. Nevertheless, gait studies continue to implement one-segment foot models to assess the effect of total ankle replacement.

Research question: The objective of this pilot study was to investigate the effect of the foot modeling approach (one-segment versus multi-segment) on how total ankle replacement is estimated to benefit or degrade the patient’s biomechanical performance.

Methods: Ten subjects with post-traumatic ankle osteoarthritis scheduled for total ankle replacement and 10 asymptomatic subjects were recruited. A one-segment and a multi-segment foot model were used to calculate intrinsic foot joints kinematics and kinetics during gait. A linear mixed model was used to investigate the effect of the foot model on ankle joint kinematic and kinetic analysis and the effect of total ankle replacement.

Results: Differences in range of motion due to the foot model effect were significant for all the gait subphases of interest except for midstance. Peak power generation was significantly overestimated when computed with the one-segment foot model. Ankle and shank-calcaneus joint dorsi-/plantarflexion range of motion did not increase post-operatively except during the loading response phase. A significant ‘group’ effect was found for stance and pre-swing phase range of motion, with total ankle replacement patients showing lower range of motion values than controls for dorsi/plantarflexion.

Significance: The outcome of this study showed that the ‘foot model’ had a significant effect on estimates of range of motion and power generation. The findings in our study therefore emphasize the clinical interest of multi-segment foot modeling when assessing the outcome of a therapeutic intervention.

Keywords: Ankle; Osteoarthritis; Prosthesis; Gait; Range of motion, Joint power

1. Introduction

End-stage ankle osteoarthritis is a common chronic disorder characterized by progressive loss of cartilage, severe physical pain and disability [1]. The current surgical treatments for end-stage ankle osteoarthritis include ankle arthrodesis and total ankle replacement (TAR). Ankle arthrodesis often relieves pain and improves quality of life scores, but may also lead to functional limitations in work and leisure activities and to adjacent joint degeneration [2,3]. Therefore, the use of motion-sparing procedures such as TAR has been encouraged, to reduce pain, to preserve the residual pre-operative ankle range of motion (RoM) to allow better function, and to decrease the risk of developing adjacent joint degeneration [4,5].

Currently, therapeutic success after TAR is primarily assessed on clinical, radiographic and questionnaire outcomes [3,6]. Recent publications have indicated that function is not

systematically restored, even if clinical and radiographic parameters show improvement [3,7]. Therefore, substantial efforts have gone into characterizing the impact of TAR on ankle joint angles and moments during gait [1,8]. A recent systematic review and meta-analysis showed that, implantation of a total ankle prosthesis increased ankle joint RoM by 3.2 degrees compared to preoperative values, but that it remained impaired compared to asymptomatic subjects [1].

Over the last decade, three-dimensional multi-segment kinematic foot models have become available for clinical use and have clearly shown their clinical relevance in detecting intrinsic foot mobility impairments [9]. Recently, multi-segment kinetic foot models have received increasing attention in methodological and clinical studies, providing new insights into the individual power distributions of the intrinsic foot joints [10–12]. These kinetic foot models further highlighted the shortcomings of computing ankle joint moments and power with a one-segment foot modeling approach, as it overestimates ankle joint power [12–14]. These shortcomings may further lead to clinical misinterpretation of how a therapeutic intervention benefits or degrades biomechanical performance, as the estimated changes simply reflect methodological errors inherent to conventionally modeling the foot as a single segment [12,15]. Nevertheless, gait analysis studies generally continue to implement a one-segment foot modeling approach to assess the effect of TAR on ankle joint angles and moments [16,17].

The purpose of this pilot study was to investigate the effect of the foot modeling approach on how TAR is estimated to benefit or degrade the patient's biomechanical performance. We compared the effect of TAR on ankle joint angles, moments and power calculated from the one-segment versus multi-segment kinematic and kinetic foot models. We also assessed the effects of walking speed, foot model and surgical intervention. In supplement, the post-operative ankle kinematics and kinetics of the patients were compared to a peer-matched control group. It was hypothesized that the outcome after TAR in ankle joint angles,

moments and power is overestimated on a one-segment compared to a multi-segment foot model (first hypothesis). Furthermore, we hypothesized that ankle joint angles, moments and power increase after TAR but remain different from the control group, irrespective of the foot modeling approach (second hypothesis).

2. Methods

2.1. Participants

A sample of 10 asymptomatic control subjects and 10 symptomatic patients participated in the study (Table 1). All patients suffered from post-traumatic ankle osteoarthritis scheduled for primary TAR between January 2017 and March 2018. Severity of ankle osteoarthritis was scored using the Canadian Orthopaedic Foot & Ankle Society (COFAS) classification system (Table 1)[18]. The inclusion criteria for the pathologic group were: post-traumatic end-stage ankle osteoarthritis with indication for TAR; exclusion criteria comprised: (1) age <18 years, (2) systemic or neurological disease, and (3) any medical problem other than post-traumatic ankle osteoarthritis liable to affect gait. All patients were scheduled for TAR within two weeks of their preoperative data collection, and were tested again 1 year after surgery. A two-component fixed-bearing Cadence® prosthesis (Integra Life Sciences, Plainsboro, NJ, USA) was implanted in all patients. Control subjects were peer-matched according to demographics; exclusion criteria comprised any medical problem possibly affecting normal gait. The local review board approved the study (B200-2017-061) and all participants signed an informed consent form.

2.2. Data collection

The measurement session consisted in asking participants to walk at a self-selected speed over a 10m walkway in which a Footscan® pressure plate (0.5m x 0.4m, 4,096 sensors, 2.8

sensors per cm²; RSscan International, Paal, Belgium) was mounted on a AMTI-force plate (0.5 x 0.4m; Advanced Mechanical Technology, Inc., Watertown, MA, USA) custom-made to fit the surface dimensions of the pressure plate. This set-up provided continuous calibration of the pressure plate with respect to the force plate, using an RsScan® 3D box. Sixteen 8mm retro-reflective markers were placed on the foot and shank of each participant according to the multi-segment Rizzoli foot model [19]. A passive optoelectronic motion analysis system (Qualysis, Göteborg Sweden) composed of eight Miquis cameras tracked kinematic data during walking over the instrumented walkway. The global acquisition frequency was 200Hz. Patients suffering from ankle osteoarthritis often experience pain during barefoot walking; to avoid maladaptive walking strategies, three representative trials were collected per participant, a trial being considered representative when the foot of interest made clear contact with the pressure plate without visual adjustments in walking behavior. Walking speed was required to remain similar across all trials in a given recording session.

2.3.Data analysis

Ankle joint angles and moments were calculated from the Rizzoli one-segment foot model (hereafter referred as the ankle joint)[20] and the IOR-4Segment-model multi-segment foot model (hereafter referred as shank-calcaneus joint) [10]. In this study, the term joint signifies the modeled biomechanical interaction between two body segments. In the Rizzoli one-segment foot model, the ankle joint is defined as the interaction between a shank (tibia + fibula) segment and a foot segment. In the IOR-4Segment-model multi-segment foot model, the shank-calcaneus joint refers to the interaction between a shank (tibia + fibula) segment and a calcaneus segment. The joint center of both joints was defined as the midpoint between the two malleoli markers.

Joint forces and moments were computed in the Inertial Coordinate System by a bottom-up inverse dynamic method using a Newton-Euler recursive algorithm based on a homogeneous matrix formalism during the stance phase of gait [21]. Kinematic and force data were filtered using a low-pass zero-lag 4th order Butterworth filter, with a cut-off frequency of 10 Hz. Inertia and weight parameters of each foot segment were discounted, as the inertia effects were negligible compared to external forces during stance [22]. The force plate data were distributed over each foot segment using a validated proportionality scheme [23]. Subarea per foot segment was estimated for each time frame by projecting the marker positions vertically on the sensor matrix of the pressure platform. The resulting center of pressure (CoP) of each estimated subarea was then used as the CoP for the corresponding foot segment in the inverse dynamics calculations. Joint moments were expressed in the proximal segment coordinate system. An in-house Matlab© program (The Mathworks Inc., Natick, MA, USA) was used for kinematic and kinetic computations. Joint angles and moments were expressed following ISB recommendations [10,19,24]. Joint moments and powers were normalized by subject-mass and all one-dimensional data were time-normalized to 100% of the stance phase.

The following discrete zero-dimensional variables were extracted from one-dimensional time-normalized kinematic and kinetic curves about the axes of the joint coordinate system: RoM of both ankle and shank-calcaneus joints for five gait sub-phases (stance phase, loading response, midstance, terminal stance and pre-swing), 3D peak joint moments and peak power generation and absorption. In addition, walking speed and stride length were extracted as discrete zero-dimensional spatiotemporal variables. The relevance of these biomechanical variables linked to the outcome of total ankle replacement is described in Table 2.

2.4. Statistics

First, the Shapiro-Wilk test was used to check data normality. Independent t-tests (significance threshold, $p < 0.05$) were conducted to compare demographic data between asymptomatic and symptomatic participants.

A linear mixed model was used to model the relationship of dependent variables (RoM) for each gait subphase of interest; joint moments and power variables over time (pre-intervention versus one year post-operative intervention) in patients receiving TAR for post-traumatic ankle osteoarthritis [25]. Since walking speed results in different foot kinetics, walking speed was implemented as an independent variable in the mixed linear model [26]. All results were analyzed in three ways: according to time (pre-intervention versus one year post-intervention), to foot model (one-segment versus multi-segment), and to time plus foot model, each as the main effect. To assess accurately the benefits of TAR, the significance of the interaction was calculated. Interaction allows the statistical significance of both time and foot model as the main effects to be calculated. This linear mixed model allows a better comparative analysis using the data when time is used as the main effect and when foot model is used as the main effect. It also accounts for any crossover effect as data might improve or worsen over time.

A second linear mixed model was used to model the relationship of dependent variables (RoM for each gait subphase of interest; joint moments and power variables between groups (patients versus controls) [25]. Since walking speed results in different foot kinetics, walking speed was implemented as an independent variable in the mixed linear model [26]. All results were analyzed in three ways: according to group (patients versus controls), to foot model (one-segment versus multi-segment), and to group plus foot model, each as the main effect. All statistical tests used R software, version 3.4.3. (<https://www.r-project.org/>; The R Foundation for Statistical Computing, Vienna, Austria). Considering that there were five statistical tests performed on the kinematic curves of each modelling approach, the significance threshold was

corrected to $\alpha = 0.01$ ($0.05/5$) for the kinematic data and to $\alpha = 0.025$ ($0.05/2$) for joint moments and power to control the type 1 error rate when performing multiple comparisons [27].

3. RESULTS

3.1. *Demographic and spatiotemporal data*

No significant differences between the two groups were found for age, weight, height or BMI; however, patients walked at a slower speed and with smaller stride length than the control subjects (Table 1). The spatiotemporal parameters showed a statistically significant increase after surgery, so that walking speed and stride length became comparable to that of the controls (Table 1).

3.2. *Pre- versus post-operative condition*

Walking speed effect. Subjects with higher walking speed had significantly greater adduction/abduction RoM during the stance and loading response phases and lower adduction/abduction RoM during pre-swing phase. Subjects with higher walking speed also had significantly lower inversion/eversion RoM during loading response and pre-swing. All peak moments and power variables were significantly affected by walking speed except for peak dorsiflexion and adduction moments. Further analysis showed that only the peak inversion moment was reduced with higher walking speed.

Time as main effect. A significant increase in RoM was observed during loading response ($P < 0.001$) for Dorsi/Plantarflexion (Table 3).

Foot model as main effect. A significant foot model effect was found for all three planes of motion during loading response, where dorsi-/plantarflexion and inversion/eversion RoM were underestimated and adduction/abduction RoM overestimated when computed with the one-segment foot model (Table 3). Differences due to the foot model were also significant for adduction/abduction RoM during stance and terminal stance and for dorsi-/plantarflexion RoM

during pre-swing, which were overestimated when computed with the one-segment foot model. Power generation was significantly overestimated when computed with the one-segment foot model.

Time plus foot model as main effect. No significant ‘time plus foot model’ effect was found for any of the investigated variables (Table 3).

3.3. Control subjects versus patients after total ankle replacement

Walking speed effect. Subjects with higher walking speed had significantly lower inversion/eversion RoM during loading response and greater dorsi-/plantarflexion RoM during terminal stance. Peak plantarflexion and abduction moments as well as both power variables were significantly increased with higher walking speed (Table 4).

Group as main effect. A significant ‘group’ effect was found for stance phase and pre-swing RoM, with TAR patients showing lower RoM values than controls for Dorsi/Plantarflexion (Table 4).

Foot model as main effect. Differences in RoM due to the foot model were significant for all the gait subphases and RoM were overestimated when computed with the one-segment foot model except for inversion/eversion RoM during loading response and terminal stance (Table 4).

A significant foot model effect was also found for peak plantar flexion moment and peak eversion moment. Both power variables were significantly overestimated when computed with the one-segment foot model.

Group plus foot model as main effect. No significant ‘group plus foot model’ effect was found for any of the investigated variables (Table 4).

4. Discussion

The aim of this pilot study was to investigate the effect of the foot modeling approach on how TAR is estimated to benefit or degrade the patient's biomechanical performance. The outcome of this study showed that the 'foot model' had a significant effect on RoM and power generation estimates. This is in accordance with earlier studies highlighting overestimation of ankle joint angles and peak power generation by the one-segment foot model compared to the multi-segment foot model [12,13,15]. The problem of modeling the foot as a single segment is that motion occurring in the intrinsic foot joints may add an extra rotation of the foot segment relative to the shank. This could then overestimate ankle joint kinematics and angular velocity, resulting in overestimation of ankle joint power [12]. However, no previous research investigated the effect of the foot modeling approach on how TAR is estimated to affect a patient's biomechanical performance. In this study, a significant pre- to post-operative increase in dorsi-/plantarflexion RoM during loading response was found for both modeling approaches (Table 3). While progress after surgery in absolute values is relatively small in degrees, it nevertheless represents a 15.4% increase in RoM for the ankle joint and a 36.8% increase in RoM for the shank-calcaneus joint. The important contribution of three-dimensional gait analysis is to monitor patients over time and specifically to assess their progress after surgery. Percentage progress in RoM is underestimated when computed with the one-segment foot model. Therefore, clinical decision-makers need to be cautious when assessing surgical interventions based on ankle joint kinematics computed with a one-segment kinetic foot model.

Differences in RoM between both foot models are not limited to dorsi-/plantarflexion, but were also found for other joint axes, especially during loading response. RoM computed with the one-segment foot model significantly underestimated inversion/eversion and overestimated adduction/abduction RoM in osteoarthritic ankles as well as in healthy ankles (Tables 3 and 4). This subphase of the gait cycle is mainly characterized by eversion of the shank-calcaneus joint associated with abduction of the calcaneus-midfoot joints. Therefore, the observed differences

in RoM may result in the segment X-axis definitions, as the long axis of the foot segment (one-segment foot model) points in a more lateral direction than that of calcaneus segment (multi-segment foot model).

The second hypothesis was only partially confirmed by the results, as the ankle and shank-calcaneus joint angles did not increase post-operatively, except for dorsi/plantarflexion RoM during the loading response. The challenge of this sub-phase of the stance phase is the abrupt transfer of body weight onto a limb that has just finished swinging forward. Therefore, a possible explanation for this might be that patients were not afraid to land properly on their ankle joint during walking. The significant increase in spatiotemporal variables seems to support these arguments, as they provide a good indicator for greater sense of stability and decreased joint pain while walking after surgery [28]. Another clinically relevant finding is that the aim of the prosthesis to preserve the residual pre-operative ankle RoM was achieved after surgery (Table 3). However, there seems to be a general agreement in earlier studies, as well as in the present results, that dorsi-/plantarflexion RoM after TAR remains impaired compared with control subjects.

No significant increase in peak plantarflexion moment and power generation was found postoperatively compared to earlier findings [16,28,29]. This discrepancy may be explained by the fact that previous studies did not include walking speed as an independent variable in their statistical analyses [16,28,29], although it is known to affect the kinetics. Although no significant difference could be demonstrated on the present data, there was a trend toward increase in both variables, which seemed from our data to be mainly due to patients' ability to walk faster, suggesting less pain after surgery. It is, however, not possible to determine conclusively to what extent the changes in muscle function leading to this trend in power generation are attributable to pain relief and to what extent they are related to improvement in biomechanical function.

5. Limitations

The findings of this pilot study should be considered in the context of four limitations. Firstly, sample size was limited to 10 symptomatic and 10 asymptomatic subjects. To detect a minimal clinical change between pre- and post-operative conditions for the variables of interest in the ankle joint (RoM, moments and power variables), 35 participants would be needed to detect a significant difference with 80% power and a p-value at 0.05. A second limitation was the partitioning of the total ground reaction forces acting on the calcaneus segment based on a proportionality scheme which combined pressure and force plate data together with marker placement data. Even though this proportionality scheme was validated, the reported ankle joint moments and power should be regarded as estimates. A third limitation was the use of skin markers to estimate joint centers and segment kinematics. Kessler et al. (2019) compared foot motion measured by biplanar videoradiography and optical motion capture and found soft-tissue artefacts of 3.29 mm on the surface of the foot [30]. However, they found also strong agreement between the two systems for foot motion in the sagittal plane which is the main anatomical plane of interest of this pilot study. The impact of these soft tissue errors on the estimation of the foot joint moments and powers is difficult to determine. Therefore, the results of the current study should be considered as an estimation and further research is needed. A fourth limitation was that the kinetics of the intrinsic foot joints were not investigated for TAR outcome. In addition, to preserve residual preoperative ankle RoM, a second biomechanical rationale for TAR rather than ankle arthrodesis is to decrease the risk of degenerative effects on the intrinsic foot joints, although this effect has not yet been proven [5]. Therefore, additional research during postoperative follow-up to monitor potential compensatory mechanisms in joints adjacent to and distal to the ankle is mandatory.

6. Conclusion

This paper provides a first estimation of the effect of the foot modeling approach on how a surgical intervention is estimated to affect a patient's biomechanical performance. Results showed that the choice of foot model had a significant effect on RoM and power generation estimates. The findings in our study therefore emphasize the clinical relevance of multi-segment foot modeling when assessing the outcome of a therapeutic intervention.

CONTRIBUTIONS

PAD conceived the original idea of the study, contributed to gait data collection, analysis and interpretation, drafted the manuscript and approved its final version.

AN contributed to gait data collection, analysis and interpretation, performed statistical analysis, provided critical revision of the manuscript and approved its final version.

TL conceived the original idea of the study, contributed to patient's enrollment and examination, provided critical revision of the manuscript and approved its final version.

RD contributed to gait data analysis and interpretation, provided critical revision of the manuscript and approved its final version.

BDB contributed to patient's enrollment and examination, provided critical revision of the manuscript and approved its final version.

JLB contributed to patient's enrollment and examination, provided critical revision of the manuscript and approved its final version.

XC contributed to gait data analysis and interpretation, provided critical revision of the manuscript and approved its final version.

LC conceived the original idea of the study, contributed to gait data analysis and interpretation, provided critical revision of the manuscript and approved its final version.

COMPETING INTEREST STATEMENT

The authors do not have any significant conflicts of interest relevant to this manuscript.

FUNDING

This study was funded in part by a clinical research Grant (BDB 2017/12) of the Belgische Vereniging voor Orthopedie en Traumatologie (BVOT). Sponsor had no involvement in this article.

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Table 1: Demographic data of patients and control subjects (Abbreviations: BMI: body mass index; SD: standard deviation; N/A: not applicable; preop: preoperative; postop: postoperative; COFAS : Canadian Orthopaedic Foot & Ankle Society ankle osteoarthritis classification¹⁴)

	Patients		Control subjects		P-values	
	Mean	SD	Mean	SD	Patients vs Control subjects	Preop vs Postop
Age (years)	62.7	8.1	61.9	6.4	0.810	N/A
Height (m)	1.7	0.1	1.7	0.1	0.928	N/A
Weight (kg)	80.5	14.2	73.9	17.6	0.370	N/A
BMI (kg/m ²)	27.6	4.3	25.2	4.4	0.237	N/A
Male:Female	4	6	6	4	N/A	N/A
Preop Walking speed (m/s)	0.93	0.17	1.18	0.13	0.001	0.009
Postop Walking speed (m/s)	1.08	0.17	1.18	0.13	0.134	
Preop Stride length (% Height)	0.66	0.06	0.76	0.06	0.001	0.002
Postop Stride length (% Height)	0.74	0.06	0.76	0.06	0.430	
COFAS						
Type 1	2					
Type 2	1					
Type 3	5					
Type 4	2					

Table 2: Biomechanical markers of human motor performance related to the assessment of the outcome of total ankle replacement

Gait variables	Rationale
Ankle RoM (degrees)	The aim of total ankle replacement is to preserve the residual pre-operative ankle range of motion. Kinematic parameters (RoM) during various subphases of the stance phase were calculated to assess if the residual pre-operative ankle RoM was preserved or increased after total ankle placement. The preservation of the sagittal ankle joint RoM is mandatory as it controls the forward movement of the body when the foot remains flat on the ground and the shank advances during the midstance phase [5].
Sub-phases of the gait cycle	Quantification of segmental RoM during specific sub-phases of the stance phase is an approach to assess if the functional objectives at the ankle and shank-calcaneus joints were accomplished for each sub-phase.
- Stance phase	The phase begins with initial floor contact and ends with ipsilateral toe-off.
- Loading Response	The phase begins with initial floor contact and continues until the other foot is lifted for swing.
- Mid Stance	It begins as the other foot is lifted and continues until body weight is aligned over the forefoot.
- Terminal Stance	It begins with heel rise and continues until the other foot strikes the ground.
- Pre-Swing	It begins with initial contact of the opposite limb and ends with ipsilateral toe-off.
Joint moments (N.m/kg)	Joint moments are considered to be primarily caused by muscles when the joint is not near the end of range of motion [24].
Power (W/kg)	Ankle power generation was reported to be a good indicator of an improvement after total ankle replacement in the ability to use the foot to propel forward and an increase in the strength of the calf muscles [4].

Walking speed (m/s) & Stride length (% Height)	Walking speed and stride length were found to reflect a greater sense of stability and decreased joint pain while walking after total ankle replacement [28].
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Table 3: Kinematic and kinetic comparison of ankle joint outcome variables measured with a 1-segment and a multi-segment foot model between pre- and post-operative conditions. *P*-values for kinematic variables ($\alpha = 0.01$) and for kinetic variables ($\alpha = 0.025$) of between-times effect, between-models effect and time*foot model interaction effect are presented. Bold *P*-values indicate significant difference.

Variables	Pre-operative (n=10) (Mean \pm SD)		Post-operative (n=10) (Mean \pm SD)		Effect (P-values)			
	One-segment	Multi-segment	One-segment	Multi-segment	Walking speed	Time	Foot Model	Time*Foot Model
Ankle RoM (°)								
- Stance								
DF/PF	17.1 \pm 5.4	13.5 \pm 5.3	21.2 \pm 4.0	16.7 \pm 2.9	0.217	0.250	0.010	0.708
INV/EV	5.3 \pm 1.9	7.0 \pm 3.4	6.5 \pm 2.0	6.9 \pm 3.9	0.076	0.625	0.761	0.456
ADD/ABD	8.2 \pm 3.2	4.0 \pm 1.1	8.7 \pm 2.4	4.11 \pm 1.5	0.032	0.419	<0.001	0.819
- Loading response								
DF/PF	6.5 \pm 2.2	6.8 \pm 2.9	7.5 \pm 2.4	9.3 \pm 2.6	0.902	0.001	0.005	0.076
INV/EV	2.6 \pm 1.1	4.4 \pm 3.1	2.1 \pm 1.6	5.3 \pm 3.6	0.008	0.075	0.003	0.306
ADD/ABD	5.2 \pm 2.9	2.3 \pm 0.7	5.5 \pm 3.0	1.6 \pm 0.6	0.003	0.865	<0.001	0.361
- Midstance								
DF/PF	8.6 \pm 3.0	6.3 \pm 2.9	10.4 \pm 2.4	8.7 \pm 2.5	0.082	0.102	0.056	0.559
INV/EV	1.8 \pm 1.1	2.9 \pm 1.2	1.9 \pm 0.7	2.4 \pm 1.1	0.356	0.223	0.279	0.400
ADD/ABD	2.3 \pm 0.9	1.9 \pm 1.0	2.8 \pm 1.2	1.9 \pm 0.9	0.126	0.493	0.023	0.377
- Terminal stance								
DF/PF	0.7 \pm 0.5	0.7 \pm 0.7	1.2 \pm 0.8	0.9 \pm 0.8	0.308	0.779	0.192	0.308
INV/EV	0.9 \pm 0.9	1.1 \pm 1.1	0.8 \pm 0.8	1.4 \pm 1.1	0.322	0.596	0.057	0.345
ADD/ABD	1.7 \pm 1.3	0.5 \pm 0.6	2.1 \pm 1.8	0.6 \pm 0.5	0.645	0.901	<0.001	0.556

- Pre-Swing RoM									
DF/PF	16.3 ± 5.4	11.9 ± 5.2	20.5 ± 4.0	15.5 ± 3.3	0.170	0.192	0.003	0.764	
INV/EV	3.6 ± 1.8	4.6 ± 2.5	4.9 ± 1.3	4.1 ± 3.4	0.021	0.822	0.401	0.190	
ADD/ABD	5.8 ± 3.9	1.9 ± 1.0	5.1 ± 2.5	2.8 ± 1.0	0.004	0.073	0.024	0.268	
Moment (N.m/kg)									
Peak DF	0.09 ± 0.04	0.07 ± 0.03	0.11 ± 0.03	0.09 ± 0.04	0.177	0.598	0.150	0.810	
Peak PF	1.20 ± 0.21	1.20 ± 0.22	1.32 ± 0.23	1.33 ± 0.24	<0.001	0.579	0.713	0.668	
Peak EV	0.09 ± 0.08	0.09 ± 0.08	0.14 ± 0.09	0.14 ± 0.09	0.013	0.486	0.911	0.983	
Peak INV	0.05 ± 0.06	0.05 ± 0.06	0.05 ± 0.03	0.05 ± 0.03	0.008	0.182	0.869	0.958	
Peak ADD	0.01 ± 0.001	0.01 ± 0.01	0.01 ± 0.01	0.01 ± 0.01	0.651	0.888	0.511	0.890	
Peak ABD	0.11 ± 0.06	0.11 ± 0.06	0.13 ± 0.07	0.13 ± 0.07	0.041	0.557	0.914	0.991	
Power (W/kg)									
Peak Absorption	0.5 ± 0.2	0.4 ± 0.2	0.6 ± 0.2	0.5 ± 0.2	<0.001	0.529	0.032	0.969	
Peak Generation	1.2 ± 0.5	0.8 ± 0.3	1.8 ± 0.7	1.3 ± 0.7	<0.001	0.335	0.006	0.829	

Table 4: Kinematic and kinetic comparison of ankle joint outcome variables measured with a 1-segment and a multi-segment foot model in control subjects and patients after total ankle replacement. *P*-values for kinematic variables ($\alpha = 0.01$) and for kinetic variables ($\alpha = 0.025$) of between-times effect, between-models effect and time*foot model interaction effect are presented. Bold *P*-values indicate significant difference.

Variables	Control subjects (n=10) (Mean \pm SD)		Post-operative (n=10) (Mean \pm SD)		Effect (P-values)			
	One-segment	Multi-segment	One-segment	Multi-segment	Walking speed	Group	Foot Model	Group*Foot Model
Ankle RoM (°)								
- Stance								
DF/PF	31.0 \pm 7.0	26.9 \pm 6.3	21.2 \pm 4.0	16.7 \pm 2.9	0.111	0.002	<0.001	0.678
INV/EV	10.4 \pm 2.6	9.2 \pm 3.0	6.5 \pm 2.0	6.9 \pm 3.9	0.214	0.057	0.366	0.395
ADD/ABD	11.8 \pm 2.3	5.4 \pm 1.7	8.7 \pm 2.4	4.11 \pm 1.5	0.622	0.176	<0.001	0.127
- Loading response								
DF/PF	7.1 \pm 1.8	7.8 \pm 2.5	7.5 \pm 2.4	9.3 \pm 2.6	0.125	0.076	0.101	0.072
INV/EV	2.9 \pm 0.8	6.7 \pm 1.3	2.1 \pm 1.6	5.3 \pm 3.6	0.049	0.068	<0.001	0.703
ADD/ABD	6.4 \pm 1.7	1.5 \pm 0.7	5.5 \pm 3.0	1.6 \pm 0.6	0.215	0.847	<0.001	0.355
- Midstance								
DF/PF	11.6 \pm 2.9	10.0 \pm 3.3	10.4 \pm 2.4	8.7 \pm 2.5	0.307	0.537	<0.001	0.809
INV/EV	2.1 \pm 1.2	2.2 \pm 0.8	1.9 \pm 0.7	2.4 \pm 1.1	0.180	0.395	0.767	0.488
ADD/ABD	2.9 \pm 2.0	1.7 \pm 0.9	2.8 \pm 1.2	1.9 \pm 0.9	0.181	0.512	0.016	0.689
- Terminal stance								
DF/PF	2.0 \pm 1.1	1.7 \pm 0.8	1.2 \pm 0.8	0.9 \pm 0.8	0.030	0.263	0.093	0.873
INV/EV	1.1 \pm 0.6	2.3 \pm 1.0	0.8 \pm 0.8	1.4 \pm 1.1	0.511	0.072	0.002	0.241

ADD/ABD	3.5 ± 1.2	0.5 ± 0.3	2.1 ± 1.8	0.6 ± 0.5	0.510	0.730	<0.001	0.019
- Pre-Swing RoM								
DF/PF	29.6 ± 6.4	25.5 ± 5.8	20.5 ± 4.0	15.5 ± 3.3	0.174	0.002	<0.001	0.296
INV/EV	9.1 ± 2.6	4.7 ± 3.0	4.9 ± 1.3	4.1 ± 3.4	0.138	0.362	0.001	0.041
ADD/ABD	7.5 ± 2.6	4.8 ± 1.9	5.1 ± 2.5	2.8 ± 1.0	0.297	0.032	0.004	0.700
Moment (N.m/kg)								
Peak DF	0.11 ± 0.05	0.09 ± 0.04	0.11 ± 0.03	0.09 ± 0.04	0.180	0.819	0.101	0.790
Peak PF	1.51 ± 0.08	1.53 ± 0.07	1.32 ± 0.23	1.33 ± 0.24	0.001	0.086	0.001	0.404
Peak EV	0.02 ± 0.09	0.02 ± 0.08	0.14 ± 0.09	0.14 ± 0.09	0.090	0.241	<0.001	0.026
Peak INV	0.06 ± 0.03	0.05 ± 0.02	0.05 ± 0.03	0.05 ± 0.03	0.681	0.651	0.120	0.495
Peak ADD	0.02 ± 0.01	0.02 ± 0.02	0.01 ± 0.01	0.01 ± 0.01	0.297	0.111	0.722	0.627
Peak ABD	0.20 ± 0.03	0.20 ± 0.03	0.13 ± 0.07	0.13 ± 0.07	0.028	0.842	0.301	0.124
Power (W/kg)								
Peak Absorption	0.82 ± 0.34	0.67 ± 0.33	0.6 ± 0.2	0.5 ± 0.2	0.038	0.454	0.015	0.645
Peak Generation	.78 ± 0.72	2.19 ± 0.72	1.8 ± 0.7	1.3 ± 0.7	<0.001	0.057	<0.001	0.378