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Assessment of the effect of a total contact cast on lower limb kinematics and joint loading

Ilias Theodorakos [a,*], Aoife Healy [b], Panagiotis Chatzistergos [b], Michael Skipper Andersen [a], Nachiappan Chockalingam [b]

[a] Department of Materials and Production, Aalborg University, Aalborg, Denmark
[b] Centre for Biomechanics and Rehabilitation Technologies, Staffordshire University, Stoke-on-Trent, United Kingdom

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ABSTRACT

Background: Total contact casts (TCCs) are used to immobilize and unload the foot and ankle for the rehabilitation of ankle fractures and for the management of diabetic foot complications. The kinematic restrictions imposed by TCCs to the foot and ankle also change knee and hip kinematics, however, these changes have not been quantified before. High joint loading is associated with discomfort and increased risk for injuries. To assess joint loading, the effect of the muscle forces acting on each joint must also be considered. This challenge can be overcome with the help of musculoskeletal modelling.
Research question: How does a TCC affect lower extremity joint loading?
Methods: Twelve healthy participants performed gait trials with and without a TCC. Kinematic and kinetic recordings served as input to subject-specific musculoskeletal models that enabled the computation of joint angles and loading. Cast-leg interaction was modelled by means of reaction forces between a rigid, zero-mass cast segment and the segments of the lower extremity.
Results: and Significance: Reduced ankle, knee and hip range of motion was observed for the TCC condition. Statistical parametric mapping indicated decreased hip abduction and flexion moments during initial contact with the TCC. The anterior knee force was significantly decreased during the mid and terminal stance and the second peak of the compressive knee force was significantly reduced for the TCC. As expected, the TCC resulted in significantly reduced ankle loading.
Significance: This study is the first to quantify the effect of a TCC on lower limb joint loading. Its results demonstrate the efficiency of a TCC in unloading the ankle joint complex without increasing the peak loads on knee and hip. Future studies should investigate whether the observed knee and hip kinematic and kinetic differences could lead to discomfort.

1. Introduction

A total contact cast (TCC) is a below the knee rigid cast which is used to immobilize and unload the foot and ankle complex. The most common applications of TCCs include the rehabilitation of ankle fractures [1] and the management of diabetic foot complications [2–4].

Studies on the effect of TCCs on the distribution of plantar loading have demonstrated that wearing a TCC substantially reduces loading at the sole of the foot [5]. Clearly, this reduction of plantar pressure will inevitably lead to a reduction in the overall loading at the ankle joint too. However, this change in ankle joint loading cannot be quantified using only measurements of plantar pressure. At the same time, the kinematic restrictions imposed by the TCC at the foot and ankle level will likely lead to compensatory changes in the kinematics of the knee and/or hip joints. Such changes could be associated with discomfort and possible injuries, especially in cases where altered kinematics contribute to an increased knee or hip joint loading. Altered knee and hip kinematics due to the immobilisation of the ankle joint have been demonstrated in the literature [6,7]. However, the actual effect of a TCC on internal joint loading has not been fully assessed. To our knowledge, there is no study in the literature that accounts for the combined effect of the TCC and muscle forces.

To achieve that, the effect of the muscle forces acting on each joint must also be considered. This challenge can be overcome with the help

* Corresponding author.
E-mail address: iliast@mp.aau.dk (I. Theodorakos).

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of musculoskeletal models of the lower limb. Such models have been used in the past to evaluate interventions such as a knee brace during gait [8], ankle braces during landing [9,10] and foot orthoses during gait [11]. These models enable the estimation of the muscle and joint reaction forces and for the first time open the way for non-invasive assessment of changes in the loading of lower extremity joints due to the use of a TCC.

Therefore, the purpose of this study was to investigate how the application of a TCC affects the kinematics and kinetics of the lower extremity during gait. We hypothesized that the TCC would (1) reduce the ankle joint range of motion (ROM), (2) reduce the ankle joint loading, (3) alter the knee and hip kinematics (4) and alter the knee and hip joint loading.

2. Methods

2.1. Participants

Twelve healthy adults (38.4 (11.5) years; 6 male, 6 female) with heights ranging from short to tall (Min: 1.57 m, Max: 1.87 m) and body mass indexes (BMI) ranging from healthy to obese (Min: 21 kg/m^2, Max: 39 kg/m^2) were recruited for this study (Table 1). People with an injury or other musculoskeletal or neurological condition that could affect their ability to ambulate unaided were excluded from this study. The study was approved by Staffordshire University's ethics committee and written consent was obtained from all individuals prior to participation.

2.2. Analysis of 3D kinematics

Two force plates (AMTI OPT464508HF sampling at 1000 Hz; AMTI, USA) and a seventeen camera Vicon motion capture system (sampling at 100 Hz; Vicon, OMG, Oxford, UK) were utilised during motion capture. Data collection consisted of two testing conditions:

1) No TCC (NoTCC): Reflective spherical markers (14 mm diameter) were placed on anatomical landmarks on the participants' pelvis and right leg using double-sided tape (Fig. 1a,b). A preliminary reference recording was done with the participants standing with their feet on the force platform. This recording was necessary to compute segments lengths and the locations of the markers in the subject-specific musculoskeletal models. Participants then walked across the laboratory at their self-selected speed and completed as many gait trials as necessary to record five acceptable trials. A recording was deemed acceptable if the participant stepped on the platform with only their right leg without visible targeting of the platform.

2) TCC: Following the barefoot walking data collection, participants had their right leg casted using Woodcast®. Woodcast® (Onbone Oy, Espoo, Finland) is a rigid casting material made from woodchips embedded in a biodegradable thermoplastic polymer [12–14]. Previous research has shown that this non-toxic and environmentally friendly material has comparable stiffness to commonly used fibreglass materials [15] and is strong enough to be safely used in weight-bearing casting applications [16] such as TCC (Fig. 1c). During casting all individual markers which were placed below the knee were removed (Fig. 1b). Following casting, reflective markers (2 marker groups, 4 markers each) were placed on the dorsal foot and anterior tibial areas of the cast (Fig. 2c). To enable the measurement of foot kinematics inside the TCC, clusters with three reflective markers were placed on the heel and the dorsum of the foot. Appropriate holes were cut in the TCC to accommodate these clusters (Fig. 1d,e). Sufficient time was allowed for the cast to harden prior to testing. Once the cast had hardened, the standardised cast shoe was fitted to the TCC (Fig. 1c). The left foot of all participants was fitted with the same make of comfort shoe. The participants were given time to familiarise themselves with walking while being in a cast prior to data collection. Similar to barefoot walking, a reference recording with the participants standing on the force plate was conducted to support musculoskeletal modelling before the recording of walking trials (five acceptable trials).

Table 1
Participant demographics. Mean and standard deviation (SD) values of participants' age, height, mass and body mass index (BMI).

Participant	Sex (M/F)	Age (year)	Height (m)	Mass (kg)	BMI (kg/m^2)
1	M	51	1.69	68.8	24
2	M	30	1.69	92.0	32
3	M	48	1.76	67.2	22
4	M	50	1.79	115.0	36
5	M	22	1.87	80.0	23
6	M	54	1.81	127.8	39
7	F	41	1.55	50.8	21
8	F	32	1.57	91.4	37
9	F	36	1.63	61.7	23
10	F	20	1.69	77.0	27
11	F	45	1.71	61.3	21
12	F	32	1.71	84.2	29
Mean:		**38.4**	**1.71**	**81.4**	**28**
SD:		**11.5**	**0.09**	**22.6**	**6.6**

2.3. Musculoskeletal modelling

2.3.1. Development of subject-specific models

Subject-specific musculoskeletal models were developed using the AnyBody Modelling System (AMS) v 7.3.4, based on the "MoCap_LowerBody" model, available in the AnyBody Managed Model Repository (AMMR) v.1.6.4. The model included a pelvis and a lower extremity comprising a femur, a tibia, a talus and a foot segment. The segments were connected with three revolute joints (subtalar, talocrural, knee) and a spherical one (hip) resulting a 12 degree of freedom (DoF) model.

The barefoot reference recordings were used to determine segment lengths for each participant, using the locations of the recorded reflective markers. Employing an optimization procedure [17] that minimized the differences between the experimental and the model markers, in a least square sense, the locations of the model markers and the scaling of the segment lengths were computed (for further details, see Lund et al. [18]). As part of this process, the length mass scaling law [19] was applied, while regression equations [20] enabled the calculation of the individual segment mass based on the total body mass.

The TCC reference recording was used to optimize the locations of the model markers on the tibia and the foot during the TCC walking, using as inputs the segment lengths as computed from the barefoot reference trial and the trajectories of the reflective markers from the TCC reference trial (Fig. 2).

For each participant, the subject-specific barefoot (NoTCC) and TCC musculoskeletal models, as obtained from the reference trials, were used for the analysis of barefoot (NoTCC) and TCC gait respectively. An optimisation procedure [21] was used to obtain the kinematics by minimizing the least square difference between modelled and experimental marker trajectories. These simulations allowed the computation of the ankle, knee and hip joint angles.

2.3.2. Cast - leg interaction

The cast was implemented into the model as a zero-mass segment, since the influence of its weight was negligible compared to the ground reaction forces. Moreover, the cast segment was assumed rigidly connected to the tibia segment. The load transfer between the lower extremity and the cast was modelled by means of reaction forces and moments between the cast and the segments of the lower extremity. Reaction forces between the cast and the tibia were assumed in the mediolateral and the anteroposterior directions while reaction moments were assumed around all axes. Additionally, a reaction force between the cast and the foot was assumed at the proximodistal direction of the tibia.

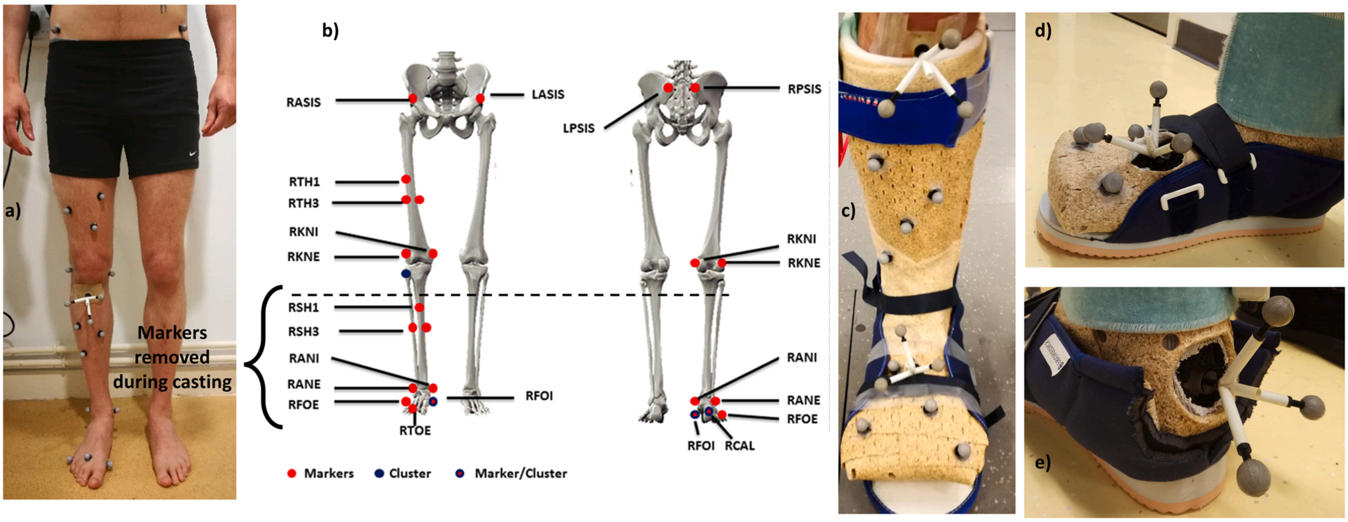

Fig. 1. Lower body marker set for NoTCC (a, b) and the changes to marker placement for TCC (c-e). c) Frontal view of a Woodcasr® TCC. d, e) The holes cut into the TCC to accommodate the two clusters.

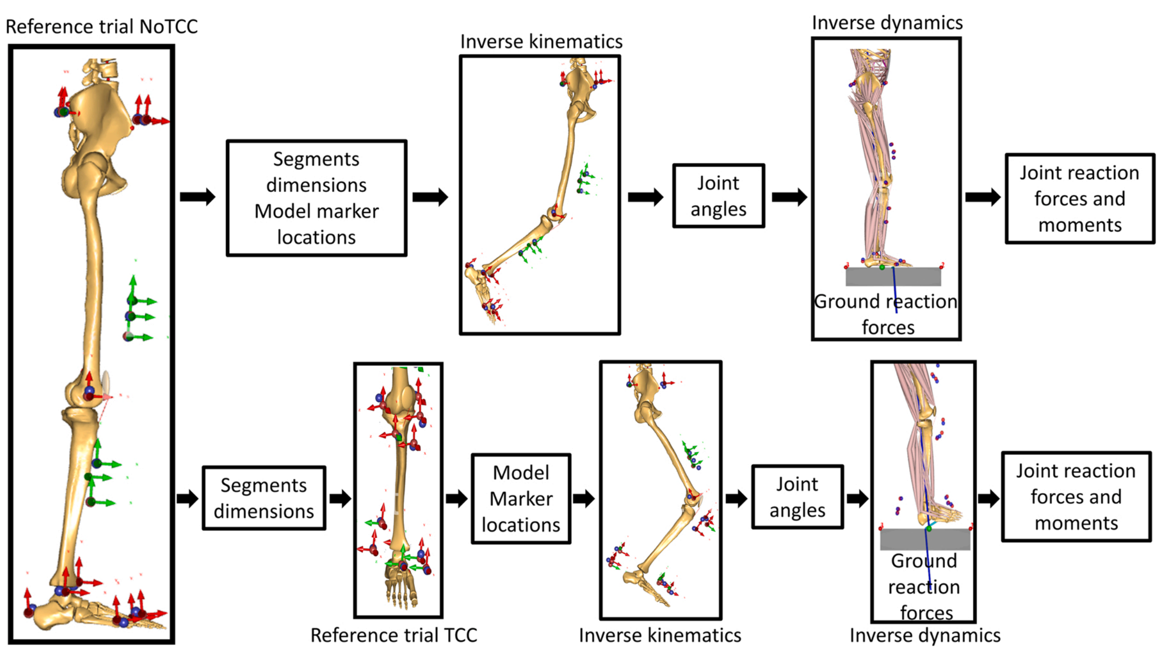

Fig. 2. Workflow of modelling procedures used to obtain joint angles, forces and moments, for the NoTCC trials (up) and TCC trials (down).

2.3.3. *Inverse dynamics*

The inverse analysis was performed using as input the ground reaction forces and marker trajectories. The muscle recruitment problem was solved using a cubic polynomial criterion, formulated as:

$$\min_{\mathbf{f}} G(\mathbf{f}^M) = \sum_{i=1}^{n^{(M)}} \left(\frac{f_i^{(M)}}{N_i^{(M)}}\right)^3 \# \tag{1}$$

Subject to.

$$\mathrm{Cf{=}d\#} \tag{2}$$

$$0 \le f_i^{(M)} \le N_i, i = 1, \ldots, n^{(M)} \# \tag{3}$$

where $f_i^{(M)}$ is the i-th muscle force which is nonnegative, meaning that muscles can only pull, $N_i^{(M)}$ is the strength of i-th muscle, $n^{(M)}$ is the number of muscles, **C** is the coefficient matrix for the dynamic equilibrium equations, **f** is a vector of unknown muscle, and joint reaction forces, and **d** contains all external loads and inertia forces (see Damsgaard et al. [22] for further details).

A constant strength muscle model was used. All simulations were batch-processed using the AnyPyTools library [23]. The estimated joint forces and joint moments were normalised to body weight (BW) and body weights times body heights (BW•BH) respectively and used for subsequent statistical analysis. The joint angles and moments were expressed in the joint reference frames as defined by Wu et al., (2002) [24], while the joint forces were expressed in reference frames fixed on the respective distal segments of the joints. The hip joint forces were expressed on the hip reference frame attached on the thigh, the knee forces on the knee reference frame fixed on the shank and the ankle forces on the ankle reference frame fixed on the foot segment. All reference frames were defined following the recommendations of the International Society of Biomechanics [24].

2.3.4. Data reduction and statistical analysis

Statistical parametric mapping (SPM) [25] was applied to test for differences in joint angles, moments and forces between conditions during the stance phase. This analysis was chosen as it allows comparison of the entire time series instead of only instances of the gait [25,26]. The SPM version of the one-way ANOVA with repeated measures was selected from the open-source code spm1d (v.0.4, spm1d.org) [25]. The analysis was performed in MATLAB (R2020A; The MathWorks, Inc., Natick, Massachusetts, USA).

Moreover, the range of motion (ROM) of the lower extremity joints were calculated for each trial. Six one-way repeated measures ANOVA tested for group differences on ankle (dorsiflexion, inversion), knee (flexion) and hip (flexion, adduction, internal rotation) ROM. The cast condition had two levels: (1) the trials performed with the cast (TCC) and (2) the trials performed without it (NoTCC). The significance level was set to 0.01 for all analyses.

3. Results

Participants 1 and 3 were excluded from the analysis due to missing experimental measurements. Average walking speed was significantly reduced (Paired samples t-test, two-tail, $t(9) = 3.603$, $p = 0.006$, $d = 1.1$) with the use of TCC from 1.29 (0.12) m/s to 0.97 (0.23) m/s.

The SPM analysis revealed statistical differences for the knee flexion and hip flexion and abduction during the early stance phase. Participants had their knee significantly more extended and their hip significantly more adducted and flexed for the casted condition (Fig. 3). Moreover, a significantly decreased ankle dorsiflexion was observed for the cast condition during the terminal stance phase (Fig. 3).

Statistically significant differences between conditions were observed for the moments that developed in the hip and for the forces developed in the knee. More specifically, hip abduction and flexion moments were decreased with the TCC (Fig. 4a). Both significant differences were observed at initial contact. With regards to joint forces, the anterior knee force was significantly decreased during the mid and terminal stance and the second peak of the compressive knee force was significantly reduced for the TCC (Fig. 4b). Moreover, the knee medial force was significantly decreased at initial contact for the TCC (Fig. 4b). As expected, ankle moments and forces were almost zero for the TCC, with the ankle inversion and dorsiflexion moments and all ankle force components being significantly decreased for the TCC (Fig. 4).

ANOVA revealed statistical differences for the ankle, knee and hip kinematics. The ankle plantarflexion ($p < 0.001$) and inversion ($p = 0.002$) ROM were significantly reduced from 23.4 (5.1)° and 14.9 (2.7)° for the NoTCC to 8.3 (3.5)° and 9.9 (2.6)°, respectively, for the TCC condition. The knee flexion ROM was significantly ($p = 0.0098$) reduced from 63.2 (7.1)° for the NoTCC to 55.0 (6.0)° for the TCC condition. Moreover, the hip internal rotation ($p < 0.001$) and adduction ($p = 0.009$) ROM were significantly reduced from 11.6 (4.0)° and 16.8 (3.5)° for the NoTCC to 8.9 (3.7)° and 12.1 (2.7)° respectively for the TCC condition (Table 2).

4. Discussion

The kinematic restrictions imposed by the TCC lead to kinematic and kinetic alterations on the lower extremity joints during gait. The presented study aimed to quantify these alterations by recruiting healthy participants, of different anthropometric characteristics, to perform gait trials with and without personalized casts made of the Woodcast® material [12–16]. The combined use of 3D kinematic analysis and subject-specific musculoskeletal modelling enabled the assessment of kinematic and, for the first time, kinetic alterations of the lower extremity joints due to the use of a TCC.

As expected, the TCC resulted in significantly reduced ankle ROM and loading. The SPM analysis also showed a consistent pattern of reduced ankle kinematics for the TCC. However, the observed statistical differences, from the SPM analysis, for the ankle joint complex were limited to the ankle dorsiflexion and only during the late stance phase, most possibly due to the high relative variability in ankle kinematics.

The hypothesis of altered knee and hip kinematics was supported by

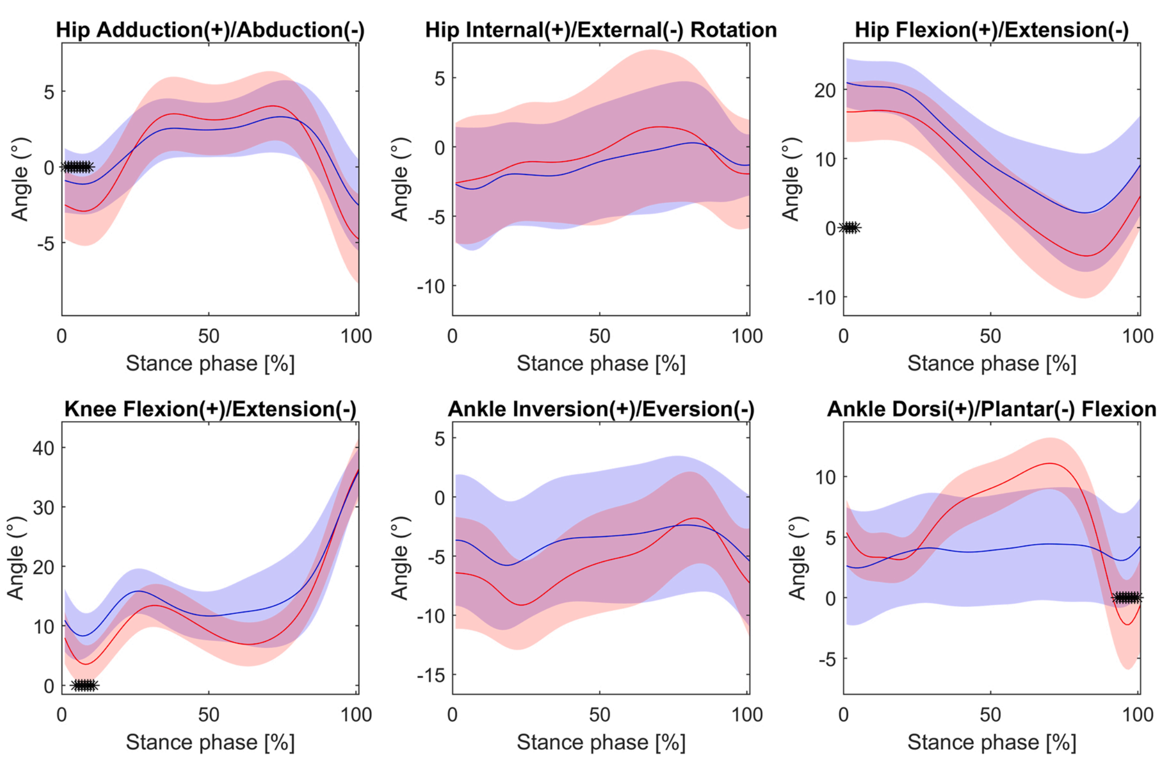

Fig. 3. Joint angles during the stance phase. The mean (± one standard deviation) values of the kinematic variables for NoTCC are illustrated in red while with blue the respective values for the trials performed with TCC. Significant differences ($p < 0.01$) are denoted with '* ' on the x axis.

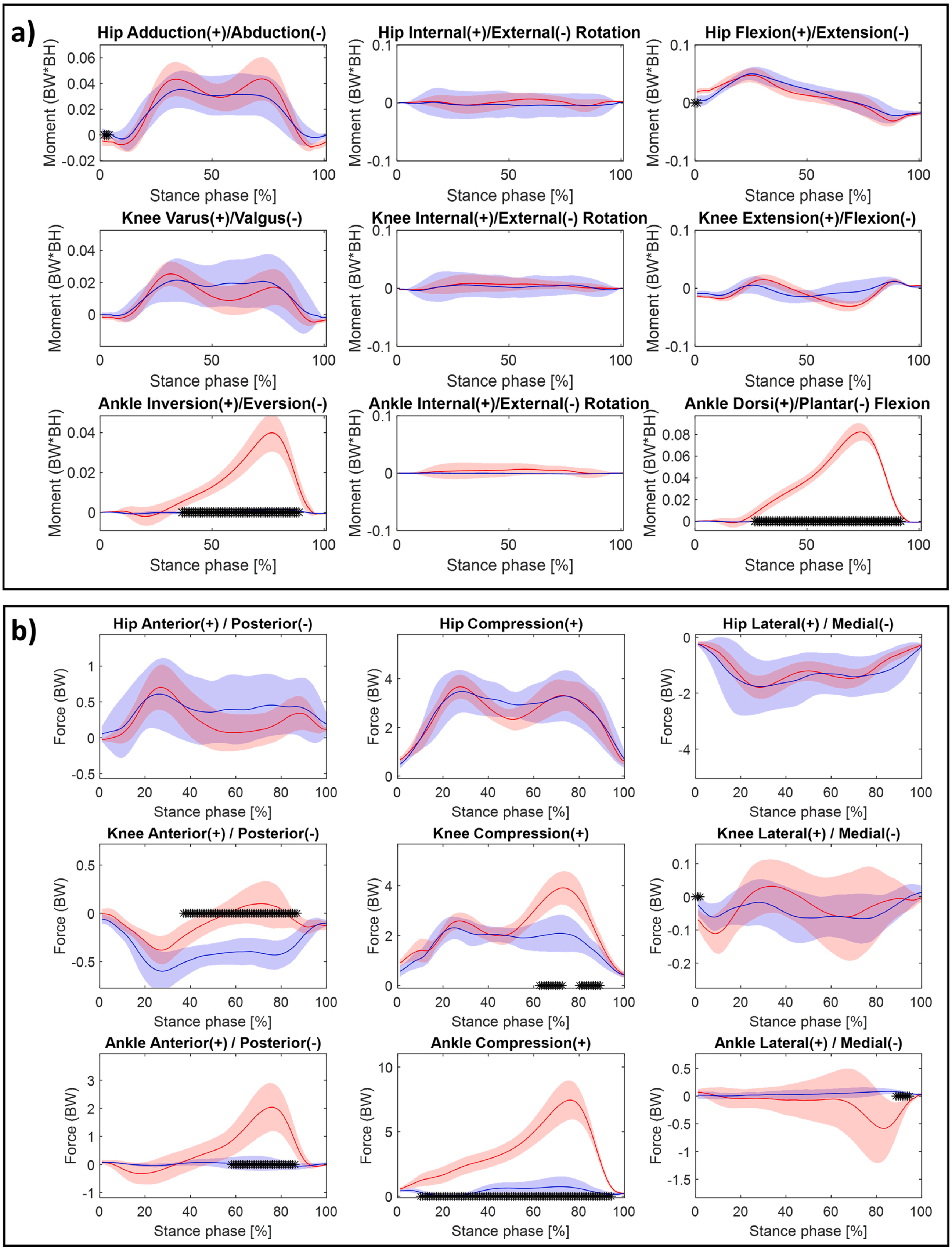

Fig. 4. Joint moments (a) and joint forces (b) during the stance phase. The mean (± one standard deviation) values of the kinematic variables for NoTCC are illustrated in red while with blue the respective values for the trials performed with TCC. Significant differences ($p < 0.01$) are denoted with '* ' on the x axis.

Table 2
Summary of ANOVA results. Significant differences ($p < 0.01$) are denoted with '* '.

ROM	Condition	Mean	(SD)°	F	p	Partial η^2
Hip Adduction*	NoTCC	16.8	3.5	10.791	0.009 *	0.545
	TCC	12.1	2.7			
Hip Internal Rotation*	NoTCC	11.6	4.0	33.953	< 0.001 *	0.790
	TCC	8.8	3.6			
Hip Flexion	NoTCC	37.8	5.8	1.952	0.196	0.178
	TCC	34.4	10.9			
Knee Flexion	NoTCC	63.2	7.1	10.663	0.010 *	0.542
	TCC	55.0	6.0			
Ankle Inversion*	NoTCC	14.8	2.7	17.204	0.002 *	0.657
	TCC	9.8	2.6			
Ankle Dorsi Flexion*	NoTCC	23.4	5.0	57.040	< 0.001 *	0.864
	TCC	8.3	3.5			

the observed significantly reduced knee and hip ROM. The participants adapted a gait pattern with less knee flexion, hip adduction and internal rotation ROM. Moreover, statistical differences were revealed for the knee and hip kinematics from the SPM analysis. For the TCC condition, participants had a more flexed knee and a more flexed and less abducted hip at and shortly after the initial contact. Similar results were reported for the knee flexion when gait with a synthetic cast was evaluated [27]. The authors reported reduced knee flexion for the most part of the stance phase, but no kinematic differences for the hip joint [27]. Differences in the reported results between the two studies can be attributed to possible differences in the stiffness of the tested casts and the anthropometric dimension of the participants. Future studies should investigate the potential implications of the observed kinematic alterations for patient comfort.

Reduced knee proximal force was observed for the TCC trials during the late stance phase. A possible explanation for this finding is the reduced activation of the ankle plantarflexor muscles at post mid-stance phase. As the cast restricts the ankle plantarflexion, the gastrocnemius and soleus muscles activity was reduced, resulting in a reduction on the observed second peak of knee proximal force that occurs at the post-mid stance phase. The observed ~1.8 BW force reduction is consistent with previous computational results [28] that showed substantial loading of the ankle plantarflexors (~1.7 BW) at post mid-stance phase. In contrast, no differences were observed on the first peak of the knee compressive force which occurs during the loading response period of the stance phase. This is explained by the fact that the thigh muscles that are the main contributors of the first peak of the knee compressive force remained unconstrained and thus, most probably, contributed in a similar manner to the knee loading between the two conditions. Similar results were observed by Stolze et al. [29] when they studied the effects of reduced ankle plantarflexors activation on the knee compressive joint force. The authors reported significant reductions on the second peak of the knee compressive force, but no differences on the first peak when they imposed reduced ankle plantarflexor muscles activity, concluding that ankle plantarflexors influence the second peak of the knee compressive force, but not the first peak [29].

Similarly, the increased posterior knee force for the casted trials during the mid and late stance phase is attributed to the reduced activity of the gastrocnemius muscle. At late stance phase with the knee being almost fully extended, the gastrocnemius generates a tibial anterior force due to the muscle activation [30]. The application of the cast restricted the action of the gastrocnemius and thus minimized the anterior force that normally would have produced on the knee joint. As a result, an increased posterior knee force was observed for the cast condition. It should be noted that the peak posterior force was statistically indifferent between the two conditions. In addition to this, the observed statistical difference for the knee medial force at initial contact was about 0.05 BW and thus, most probably clinically insignificant.

In contrast to the knee, statistical differences for the hip loading were observed only close to the initial contact. The observed decreased hip abduction moment is supported by previous results where reduced hip abduction moment was reported during the early stance phase, for gait trials performed with a synthetic cast [27]. However, our observed hip flexion moment decrease opposes previous findings where no differences were reported [27].

The fact that none of the knee and hip kinetic variables, in our study, displayed increased peaks for the TCC condition can be perceived as a positive finding. High joint loading is associated with an increased risk for injuries on the structures surrounding the joints [31]. However, as the musculoskeletal model used in the present study does not model the ligamentous structures individually, it cannot provide outputs that would allow us to investigate the distribution of the joint loading on individual ligamentous structures. Our results cannot be directly compared against studies that employed musculoskeletal models to investigate different interventions that restrict the ankle and/or different testing tasks (e.g., running, landing) [9–11]. However, a similar pattern can be identified: mechanical constrains on the ankle seem to result in kinematic and/or kinetic differences on the hip and/or knee computational variables. In the past, computationally predicted joint loading has been associated with sitting discomfort [32]. It should be investigated whether similar patterns can be identified for gait to inform the design of interventions that counteract any possible discomfort.

A small number of participants were recruited for this study. To account for this fact, a more conservative level of significance was selected (a=0.01). Moreover, the control trials were performed barefoot, despite the known differences between barefoot and shoed gait [33]. Due to lack of standardized shoes to accommodate the variability of our participants, barefoot gait was selected over performing the control trials with non-standardized footwear. Including participants with high BMI in studies involving motion capture using skin markers and musculoskeletal modelling pose specific challenges due to soft tissue artifacts. However for the present study recruiting people with high BMI was deemed necessary to account for the fact that people with diabetes also tend to have high BMI scores [34]. Another limitation of this study was the lack of quantification of the lower extremity cast interaction. This could be achieved by employing a full 6 DOF instrumented cast. Then, the obtained measurements could be implemented into the musculoskeletal model, however such instrumentation was not available for this study. The absence of walking speed control between the conditions could be regarded as a limitation of this study [35]. As a result, the findings presented here correspond to the overall effect of TCC which is expected to reduce the participants' walking speed. More research will be needed to separately quantify the effect of loading speed on joint loading and kinematics during TCC walking. Our results are limited on the intervention, population and the task presented here as different interventions lead to different ankle constrains which most probably would result in different lower extremity biomechanical behaviour, while the joint loading is also task depended.

Our results on the effect of a TCC on normal gait demonstrated its efficiency in unloading the ankle joint complex, without increasing the peak loads on knee and hip, despite the observed kinematic and kinetic knee and hip differences. These findings offer a quantitative support for the clinical use of TCCs, while future studies should evaluate whether the observed kinematic and kinetic differences on the knee and hip could contribute to patients' discomfort.

Conflict of interest statement

Onbone, Espoo, Finland provided their material for this study and partially funded this study but had no involvement in experimental design or the preparation of this manuscript.

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