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The Knee

Effects of an over-ground exoskeleton on external knee moments during stance phase of gait in healthy adults

Chris A. McGibbon [a,b,*], Scott C.E. Brandon [a,c], Marcus Brookshaw [a], Andrew Sexton [a]

[a] Institute of Biomedical Engineering, University of New Brunswick, Fredericton, New Brunswick, Canada
[b] Faculty of Kinesiology, University of New Brunswick, Fredericton, New Brunswick, Canada
[c] School of Engineering, University of Guelph, Guelph, Ontario, Canada

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ABSTRACT

Background: Physical activity and exercise is central to conservative management of knee osteoarthritis (KOA), but is often difficult for patients with KOA to maintain over the decade or more prior to surgical management. Better approaches are needed for maintaining physical function and health in this population that can also address the patho-biomechanics of the osteoarthritic knee.
The objective of the study is to quantify how a lower-extremity robotic exoskeleton (dermoskeleton) modifies the external knee moments during over-ground walking in a sample of healthy adults, and to evaluate these biomechanical modifications in the context of the osteoarthritic knee.
Method: Motion analysis data was acquired for 13 participants walking with and without the dermoskeleton. Force plate data, external knee moment arms, and knee moments in the laboratory and tibia frames of reference were computed, as well as time–distance parameters of walking, and compared between the two conditions.
Results: Although gait speed was not different, users took shorter and wider steps when walking with the dermoskeleton. Ground reaction forces and early-stance knee moment increased due to the added mass of the dermoskeleton, but the knee adduction moment was significantly reduced in late stance phase of gait. There was no effect on the knee torsional moment when measured in the anatomical frame of reference, and the late-stance knee flexion moment was invariant.
Conclusions: The dermoskeleton demonstrated favorable biomechanical modifications at the knee in healthy adults while walking. Studies are warranted to explore this technology for enabling physical activity-based interventions in patients with KOA.

1. Introduction

Osteoarthritis (OA) is a debilitating and costly disease [3] that has significant impacts on health and quality of life [4]. OA is estimated to affect more than 10% of the adult population [5], and while most prevalent in older adults, more than 1/3 of OA patients are under the age of 55 years [6]. The knee is the most commonly affected joint from OA [7], often leading to profound

Abbreviations: KOA, Knee osteoarthritis; KAM, Knee adduction moment; CoP, Center of pressure; KJC, Knee joint center; F, Ground reaction force; T, Ground reaction (twist) torque; R, External knee moment arm (lab frame); M/M′, External knee moment (lab frame/tibia frame); PoI, Parameter of interest; NON, Not wearing device; KON, Wearing device; BW, Body weight; HT, Height.

* Corresponding author at: Institute of Biomedical Engineering, University of New Brunswick, 25 Dineen Drive, Fredericton, New Brunswick E3B5A3, Canada.
E-mail address: cmcgibb@unb.ca (C.A. McGibbon).

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mobility and balance impairment [8,9], increased social isolation [10], and loss of employment income and opportunity [11]. Although total knee arthroplasty (TKA) has a high level of success in restoring mobility function in patients with KOA [12], long-term function of the operated knee does not always meet patients' expectations [13], and a significant number of KOA patient never elect to have TKA [14].

There are numerous factors that contribute to the success of TKA. Recent research suggests that patients undergoing TKA with moderate radiographic OA (Kellgren–Lawrence scale scores, KL < 3) have worse outcomes compared to those with severe OA (KL ≥ 3) [15,16]. In addition to having more severe pre-operative grade of OA, patients with better outcomes also have fewer involved joints and fewer comorbidities [17]. This creates a significant conundrum for KOA management in patients whose symptoms are severe, or for whom conservative treatments have failed, yet have not progressed radiographically to be indicated for TKA. There is considerable discordance between knee pain intensity and radiologic disease severity [18,19], and it is estimated by London et al. [14] that 3.6 million Americans are currently living with debilitating symptoms that cannot be alleviated through conservative management.

For the young KOA patient lingering in the so-called "treatment gap" the implications are even more significant [20]. Prolonged inability to exercise or remain physically active is highly likely to cause co-morbidity in later stages of the disease [21] and potentially allows mobility related neuromuscular maladaptations to form (cf. [22–25]) that can persist after TKA [26,27]. Although this argument has also been used to propose less invasive surgical implants, such as the KineSpring® [28], there remains a significant need for conservative interventions that can address the patho-biomechanics of the OA knee at both the structural and neurological level of the lower-extremity musculoskeletal system, and that can enable physical activity and exercise prescriptions to be followed and monitored by patients and clinicians.

In this paper we explore the potential of using a bilateral knee-powered lower-extremity exoskeleton, called a "dermoskeleton" (B-Temia Inc. Quebec City, QC), as a mobility assist device to enable physical activity in the KOA population, by evaluating how the technology modifies external knee moments during over-ground walking in a group of healthy young adults. Specifically, we identify the mechanism(s) responsible for the differences observed in moments (flexion/extension, abduction/adduction, internal/external rotation) applied to the knee from the foot-ground kinetics during the stance phase of gait, when users walk at their preferred pace with and without the dermoskeleton.

Based on the observed biomechanics, and the available literature of knee moment patterns in the KOA population, we evaluate if the dermoskeleton may be biomechanically indicated for KOA, and outline future studies aimed at a gaining fuller understanding of this technology for enabling and protecting the musculoskeletal system. A better understanding of this emerging technology is needed in order to determine its potential applications for mobility assistance and rehabilitation in patient populations.

2. Methods

2.1. Subjects and procedures

The study was approved by institutional ethics review board. Fifteen healthy adults (seven females; mean (S.D.) age, 27.0 (6.4) years) recruited from the local university and surrounding community participated in the study, which required two visits to the motion analysis laboratory on the university campus. All volunteers provided signed informed consent prior to participating. Volunteers were also largely naive to exoskeletons (barring general knowledge from popular culture), and none of the participants in the study had worn the dermoskeleton prior to consenting to the study.

The motion analysis laboratory used (HPL, Andrew and Marjorie McCain Human Performance Laboratory, located at the University of New Brunswick), shown in Figure 1A, was a large-volume motion capture lab integrated with an indoor running track that enabled participants to pass through the laboratory for data collection during continuous over-ground walking or jogging. Motion analysis equipment in the HPL consisted of twelve wall-mounted Vicon (T-160, 16 megapixel) optoelectronic motion tracking cameras, and six Kistler force platforms (9281EA) arranged in a 3 × 2 matrix flush with the floor surface (Figure 1A). A portable data acquisition system was carried in a backpack and used to acquire synchronous EMG, accelerometry, and signals from the dermoskeleton's internal sensors (Figure 1B).

The dermoskeleton was made by B-Temia Inc. (Quebec City, QC, Canada) and consists of bilateral motors for assisting left and right knees, a pelvis belt and chariot system for suspending the device, and thigh and shank cuffs for attaching the rigid dermoskeleton "links" (thigh beam and tibia shell) to the user. The controller recognizes gait (walking, jogging and running full speed), stair ascent and descent, chair rising, bending at the knee, and other activities of daily living. Each leg of the device is connected to the belt with a flexible joint that allows rotation in the frontal and transverse planes, and a sagittal-plane pin joint that permits unrestricted hip flexion–extension. The knee joint consists of two degrees of freedom: a sagittal plane axis actuated by the motor, and a frontal-plane pin connection allowing subject-specific knee adduction alignment. The device weighs 4.5 kg, and the backpack adds another 4.7 kg (2.9 kg for the research instrumentation shown in Figure 1B plus 1.8 kg battery). Participants wore the backpack during both unassisted (NON, without the dermoskeleton) and assisted (KON, with the dermoskeleton) conditions to eliminate its influence on comparisons.

For the first visit, participants were introduced to the dermoskeleton with a general description of how the device operates. The participant was then measured and fitted with appropriate size belts and leg cuffs, and the dermoskeleton thigh beam was adjusted to align with the knee and hip (lateral) axes of rotation with the participant's joints. The dermoskeleton was then donned by first securing the pelvis belt and chariot system (left and right sides which support one of the dermoskeleton's leg segments), and gradually increasing the snugness of the lower-leg cuffs while leaving thigh cuffs less snug, until the dermoskeleton

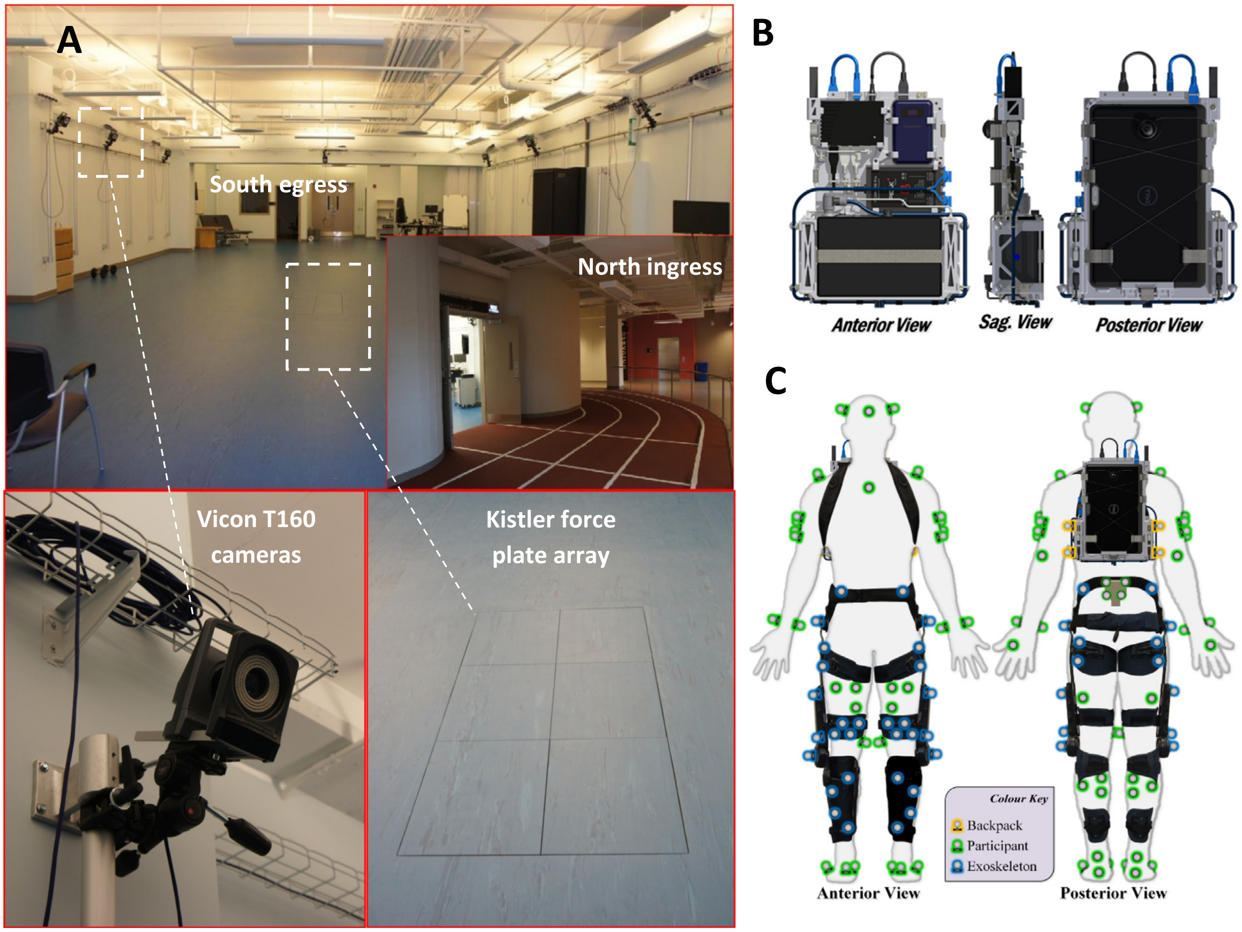

Figure 1. A) Motion analysis laboratory used in the study. Top: Laboratory viewing volume and inset showing lab entrance from 140 m indoor track. Bottom: Speed-rail mounted Vicon T160 motion cameras (12 cameras total) and six Kistler force platforms (3 × 2 arrangement). B) Custom built portable data collection system for acquiring EMG and sensors from the dermoskeleton. C) Marker set configuration used to independently track the user and dermoskeleton.

was securely suspended and allowed free movement of all of the participants' lower-limb joints. This was determined by judgment of the tester and participant feedback after moving about and practicing functional activities with the dermoskeleton motor in its passive state.

After fitting, the dermoskeleton controller was "tuned" for the participant. The dermoskeleton provides assistance during four portions of the gait cycle, as follows: i) during weight acceptance, mechanical resistance is provided to the knee to stabilize the joint, ii) through mid-stance, the device provides extension assistance to the weight-bearing leg, iii) during push-off and early-swing, flexion assistance is provided, and iv) during mid-swing, extension assistance is provided (ending before terminal swing). After initial tuning, the participant practiced various activities and any adjustments to tuning parameters and/or fitting were performed, until the tester and participant agreed the optimal fitting and tuning had been achieved. All tuning parameters for participants were stored on the tablet using a study ID #. All fitting data were also documented.

Within one week, participants returned for the second visit. They were first re-introduced to the dermoskeleton, and fitting and tuning data from visit 1 were applied, followed by a short practice of the activities to be performed in the laboratory. Next, the participant was prepared for motion analysis, which used a set of 76 reflective markers be attached to legs, pelvis, torso and arms (Figure 1C). Due to the need to accommodate the dermoskeleton and still enable comparison of the participants' biomechanics in the KON and NON conditions, the marker configuration used for the lower-extremity was non-standard, and therefore warrants some detailed discussion.

Body segment marker clusters were pre-positioned at the start of the session to allow unencumbered donning/doffing of the dermoskeleton, which itself was outfitted with a set of 34 permanent reflective markers (as shown in Figure 1B), such that each "limb segment" of the leg and device had an independent cluster of 3 to 4 markers. The dermoskeleton motor located at the knee did not allow a lateral human knee marker to be used, and because the dermoskeleton is suspended by the waist belt and chariot system, tracking the anterior superior iliac spines (ASIS) in the conventional manner of placing markers on the palpated landmarks of the ASIS is not possible in the KON condition.

A sacral plate with a cantilever for supporting a marker cluster was used to overcome this challenge. The sacral plate apparatus was a lightweight 3D printed structure that was secured to the participant's bare sacrum with double sided tape and an elastic overwrap, allowing the cantilevered marker cluster to remain in place both with and without the dermoskeleton belt being worn. Prior to donning the dermoskeleton, the participant's ASIS and lateral knee epicondyles were marked and static trials acquired in order to register the virtual ASIS (and ultimately hip centers) in sacral cluster coordinates, as described previously [29], and the lateral knee marker in thigh cluster coordinates. Knee and hip centers were determined using range of motion trials (star-arc test and SCoRe method for the hip [30], and active knee flexion/extension and SARA method [31] for the knee).

A variety of activities were completed in two blocks: one with the dermoskeleton in its active mode (KON) and one without wearing the device (NON), in randomized order. The entire testing session lasted approximately two hours. Although data were collected for static standing, sit-to-stand, deep knee bends, walking, and jogging, in this paper we only report data from the stance phase of walking. For walking trials, a minimum of four trials for each condition (KON and NON) was required with clean (one foot per plate and one plate per foot) foot strikes on the force plates defining a full stride of the right and/or left feet.

2.2. Data analysis

The goals of this study were to examine the external knee moments during stance phase of gait when walking with and without a dermoskeleton, and to identify the degree to which alterations in the ground reaction forces and external knee moment arms (from center of pressure, CoP, to anatomical knee joint center, KJC) contribute to the observed difference in external knee joint moment.

Although it was possible to conduct this as an inverse dynamic analysis using a traditional link-segment model [32], the reduction of the model to a simple knee-wrench allows us to more directly quantify the relative effects of 3D anatomical orientation/alignment (dynamic moments arms) and 3D external loading (ground reaction forces and twist moment at the CoP) on the resulting torques applied to the knee. Because we are comparing within-subjects (KON versus NON condition) and limiting the analysis to the stance phase of gait, we also ignored the effects of leg mass and inertia (quasi-static). Biomechanically, the analysis was very straight forward and is illustrated in Figure 2. Given the ground force vector $\boldsymbol{F}=[F_x, F_y, F_z]$ and twist torque $\boldsymbol{T}=$

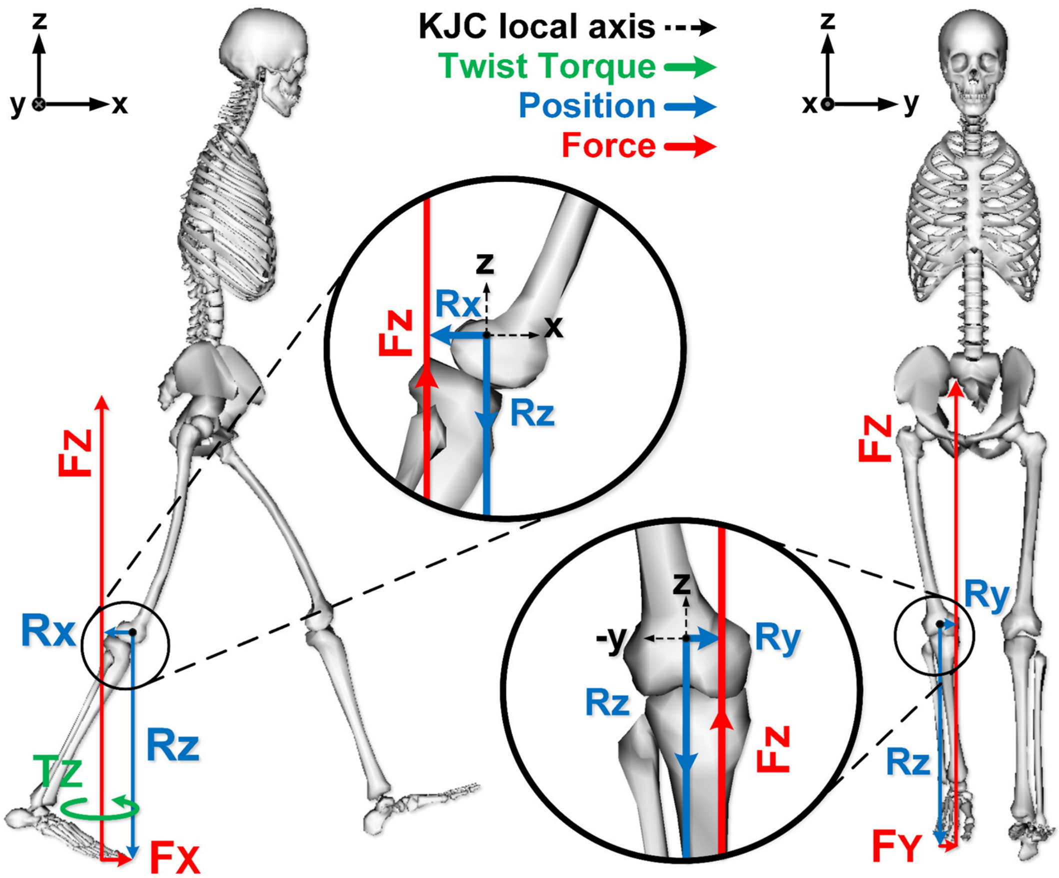

Figure 2. Illustration of the external knee moments calculated by the quasi-static model in laboratory coordinates.

$[0, 0, T_z]$,and the knee moment arms vector $\boldsymbol{R} = \mathrm{CoP} - \mathrm{KJC} = [R_x, R_y, R_z]$, the external knee moment was defined in the laboratory global reference frame as the cross-product of the moment arm and ground force vectors plus the external torques, $\boldsymbol{M} = \boldsymbol{R} \times \boldsymbol{F} + \boldsymbol{T}$.

Normalization of the data proceeded by first interpolating the kinematic and kinetic variables using a cubic spline function (1001 points) between foot strike and toe-off, to represent stance phase (0 to 100% in .1% increments). Magnitudes of kinematic variables (R, CoP) were normalized to the subject's height and multiplied by 100 (% HT). Magnitudes of ground reaction forces for both NON and KON conditions were normalized to human body weight (9.81∗body mass in kilograms, without the exoskeleton) and multiplied by 100 (% BW), and moments/torques were thus normalized to body weight and height (% BW × HT). Normalizing both NON and KON results to the individual's unladen body weight reflects the physiologically-relevant load to anatomical structures.

To account for potential effects of leg posture variability on the within-subjects comparisons, knee moments were also computed in the tibia frame of reference. An orthogonal reference frame was constructed using the ankle–knee joint center as the transverse plane axis and knee medio-lateral epicondyle markers to locate the sagittal plane axis and frontal plane axis by conventional cross-product approach. External knee moments were then expressed in this frame of reference.

Parameters of interest (PoI) were then extracted from the time- and magnitude-normalized kinematic and kinetic data, as summarized in Table 1. PoI for ground reaction force (F), foot twist torque (T), moment arms (R), and resulting moments at the knee (M) (in both the lab and tibia reference frames), were selected based on maxima and minima that occur during early, mid and late stance phase of gait. The PoI were then compared between KON and NON conditions using a paired samples t-test. CoP excursions were calculated by subtracting the CoP at heel strike, but were not included in the statistical analysis since it was the moment arm or relative location of the CoP to the KJC that was of interest.

In order to evaluate the quality of gait in the two conditions, and to potentially help explain any observed differences in the knee kinetics, time–distance parameters were also computed for walking trials. These consisted of gait velocity (average forward velocity of pelvis markers), stride length (distance between successive ipsilateral heel strikes), stride velocity (stride length over time between successive heel strikes of ipsilateral limb), step length and width (a/p and m/l distance between ipsilateral and contralateral feet during their respective mid-stance of gait), stance phase percent (stance time over cycle time), single support percent (swing time over stance time), and toe-out angle (angle of foot long-axis with respect to direction of progression). These data were similarly compared between NON and KON conditions using a paired samples t-test.

All data processing and analysis was performed with custom written Matlab (MathWorks Inc. Natick MA) programs.

3. Results

Of the fifteen participants, two were excluded due to lost markers from the exoskeleton during the subject's session which prevented accurate computation of kinematics; as such the results below include data for the remaining 13 subjects.

Table 1
Parameters of interest in kinematic and kinetic variables during the gait cycle of walking.

Parameter	Description
${}^{max}Fz_{early}$	Vertical ground force, early stance maximum
${}^{min}Fz_{mid}$	Vertical ground force, mid stance minimum
${}^{max}Fz_{late}$	Vertical ground force, late stance maximum
${}^{max}Fy_{early}$	Medio-lateral ground force, early stance maximum
${}^{min}Fy_{mid}$	Medio-lateral ground force, mid stance minimum
${}^{max}Fy_{late}$	Medio-lateral ground force, late stance maximum
${}^{min}Fx_{early}$	Anterio-posterior ground force, early stance minima
${}^{max}Fx_{late}$	Anterio-posterior ground force, late stance maxima
${}^{min}Tz_{early}$	Vertical twist torque, early stance minima
${}^{max}Tz_{late}$	Vertical twist torque, late stance maxima
Rz_{early}	Vertical knee moment arm, early stance value
Rz_{mid}	Vertical knee moment arm, mid stance value
Rz_{late}	Vertical knee moment arm, late stance value
Ry_{early}	Medio-lateral knee moment arm, early stance value
Ry_{mid}	Medio-lateral knee moment arm, mid stance value
Ry_{late}	Medio-lateral knee moment arm, late stance value
Rx_{early}	Anterio-posterior knee moment arm, early stance value
Rx_{mid}	Anterio-posterior knee moment arm, mid stance value
Rx_{late}	Anterio-posterior knee moment arm, late stance value
Mz_{early}	Knee twist moment, early stance phase
Mz_{late}	Knee twist moment, late stance phase
Mz_{term}	Knee twist moment, terminal phase
My_{early}	Knee flexion/extension moment, early stance phase
My_{late}	Knee flexion/extension moment, late stance phase
My_{term}	Knee flexion/extension moment, terminal stance phase
Mx_{early}	Knee varus/valgus moment, early stance phase
Mx_{mid}	Knee varus/valgus moment, mid stance phase
Mx_{late}	Knee varus/valgus moment, late stance phase

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Table 2
Time–distance parameters for walking with (KON) and without (NON) the dermoskeleton.

	NON		KON		Sig.
	Mean	S.D.	Mean	S.D.	*p*-Value
Gait velocity (m/s)	1.56	0.17	1.50	0.16	.110
Stride velocity (m/s)	1.54	0.16	1.50	0.16	.216
Stride length (m)	1.56	0.13	1.51	0.13	.078
Step length (m)	0.79	0.08	0.76	0.06	.018*
Step width (m)	0.13	0.04	0.15	0.03	.002*
Stance phase (% cycle)	58.16	8.00	58.48	6.33	.903
Single support (% stance)	61.40	5.02	61.06	3.50	.804
Toe-out angle (°)	3.95	1.93	4.36	1.73	.406

* Significant at $p < .05$.

3.1. Time–distance parameters

Time–distance parameters are shown in Table 2. Gait velocity (average pelvis velocity, $p = 0.110$) and stride velocity (stride length over stride time, $p = 0.216$) did not differ between KON and NON conditions. There was also no significant difference in stride length ($p = .078$), but a significant difference was observed for step length ($p = .018$) and step width ($p = .002$), where participants took slightly shorter and wider steps in the KON condition compared to the NON condition. Percent of gait in stance phase, percent of stance in single support, and toe-out angle, were not significantly different between KON and NON conditions ($p > .05$).

3.2. Biomechanical analysis of the knee

PoI statistics are shown in Tables 3 through 5 for the kinematic and kinetic events listed in Table 1. Stance phase cycled plots for each kinematic and kinetic variable are shown in Figures 3 through 7. For the first column of plots (*a*, *b*, *c*) the solid lines are sample means and shaded regions are ±1 standard deviation (SD) boundaries for the NON (blue) and KON (red) conditions. The overlay effect creates a purple region where SD boundaries overlap. The vertical dotted lines identify the PoI, and their labels indicate a * if the difference between NON and KON conditions was significant (at $p < .05$).

For the second column of plots (*d*, *e*, *f*) in Figures 3 through 7, the solid line represents the mean of the differences and shaded regions the ±1 SD boundary on the mean of the differences. The shading color is set by which testing condition had the highest magnitude of value for that variable at each % stance, where blue shaded regions indicate that the absolute magnitude for the NON condition was greater than or equal to the KON condition, and the red shaded regions indicate that the absolute magnitude for the KON condition was greater than the NON condition. This provides a simple tool for visualizing how wearing the dermoskeleton changes the kinematics and kinetics of the external force and moment couple acting on the user's knees.

3.2.1. Foot–floor interactions

The CoP anterior excursion (Figure 3*a*) was slightly lower for the KON condition compared to the NON condition throughout stance phase, resulting in a slightly smaller total excursion, and the medio-lateral excursion of the CoP (Figure 3*b*) was more laterally positioned with respect to heel strike location, and much less variable throughout most of early to mid-stance. Figure 3*d* and *e* shows the difference in excursion kinematics of the CoP between NON and KON conditions were about ±1% HT.

All of the peak magnitudes in normalized ground reaction forces (Figure 4*a*, *b*, *c*) and twist moment (Figure 3*c*) were significantly different ($p < .05$) between the NON and KON conditions. Figures 4*d*, *e*, *f* and 3*f* show respectively that in the KON

Table 3
Comparison of ground reaction force parameters of interest (PoI) during walking with (KON) and without (NON) the dermoskeleton.

PoI†	NON		KON		KON-NON		Sig.	95%CI	
	Mean	SD	Mean	SD	Mean	SD		Lower	Upper
${}^{max}Fz_{early}$	127.78	10.51	136.35	10.33	8.57	5.61	<0.001*	5.18	11.96
${}^{min}Fz_{mid}$	68.27	9.10	76.09	11.28	7.82	6.50	0.001*	3.90	11.75
${}^{max}Fz_{late}$	127.37	5.68	132.39	7.51	5.02	4.81	0.003*	2.11	7.93
${}^{max}Fy_{early}$	7.64	2.29	9.31	2.27	1.67	1.53	0.002*	0.75	2.59
${}^{min}Fy_{mid}$	2.38	1.10	3.33	1.18	0.96	0.78	0.001*	0.48	1.43
${}^{max}Fy_{late}$	5.62	2.40	6.84	2.39	1.23	1.69	0.022*	0.21	2.25
${}^{min}Fx_{early}$	−26.28	4.55	−28.27	4.99	−1.99	2.88	0.028*	−3.73	−0.25
${}^{max}Fx_{late}$	27.50	3.52	29.09	3.95	1.60	2.14	0.019*	0.31	2.89
${}^{min}Tz_{early}$	−0.13	0.15	−0.22	0.14	−0.09	0.13	0.037*	−0.16	−0.01
${}^{max}Tz_{late}$	0.40	0.17	0.62	0.16	0.22	0.17	<0.001*	0.12	0.32

† F = ground reaction force (%BW), T = ground twist moment (%BW∗HT). Laboratory coordinates: X = anterior–posterior, Y = medio-lateral, Z = vertical.
* Significant at $p < .05$.

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Table 4
Comparison of knee moment arm parameters of interest (PoI) during walking with (KON) and without (NON) the dermoskeleton.

PoI†	NON		KON		KON-NON		Sig.	95%CI	
	Mean	SD	Mean	SD	Mean	SD		Lower	Upper
Rz_{early}	−28.40	0.89	−28.41	0.84	−0.02	0.12	0.617	−0.09	0.06
Rz_{mid}	−28.58	1.00	−28.50	0.88	0.08	0.16	0.101	−0.02	0.17
Rz_{late}	−28.09	0.97	−27.83	0.90	0.26	0.17	<0.001*	0.16	0.36
Ry_{early}	1.17	0.62	0.50	0.61	−0.67	0.34	<0.001*	−0.88	−0.46
Ry_{mid}	0.64	0.98	0.27	0.85	−0.37	0.30	0.001*	−0.56	−0.19
Ry_{late}	0.58	1.10	0.17	0.98	−0.41	0.29	<0.001*	−0.59	−0.23
Rx_{early}	0.92	0.91	0.79	0.76	−0.13	0.55	0.416	−0.46	0.20
Rx_{mid}	1.01	0.72	0.35	0.75	−0.66	0.58	0.001*	−1.01	−0.31
Rx_{late}	−2.45	0.79	−2.73	0.73	−0.28	0.38	0.019*	−0.51	−0.06

† R = moment arm (%HT). Laboratory coordinates: X = anterior–posterior, Y = medio-lateral, Z = verical.
* Significant at $p < .05$.

condition the fore-aft shear forces about two percent BW higher, medio-lateral shear force two percent BW higher, vertical ground reaction forces were about five to 10% BW higher, and twist torque was 0.3% BW∗HT higher, than for the NON condition, owing naturally to the increased inertia of the body when wearing the exoskeleton. Statistical results are summarized in Table 3.

3.2.2. Knee moment arms

The medio-lateral Y axis moment arm was smaller in the KON condition throughout almost all of stance phase (Figure 5*b* and *e*), and was significantly smaller than NON moment arms in early (Ry_{early}, $p < .001$), mid- (Ry_{mid}, $p = .001$) and late (Ry_{late}, $p < .001$) stance phase. The vertical moment arm (Figure 5*c* and *f*) was only different in late stance (Rz_{late}, $p < .001$). In the anterior–posterior direction (Figure 5*a* and *d*), the moment arm in mid-stance was larger for NON condition in mid-stance (Rx_{mid}, $p = .001$) but larger for KON in late stance (Rx_{late}, $p = .019$). Statistical results are summarized in Table 4.

3.2.3. Knee external moments

Many of the external moment PoI were significantly different between the KON and NON conditions (Figures 6 and 7, and Table 5). In laboratory frame of reference, the 2nd peak in adduction moment (Figure 6*a*, My_{term}) and peak extension moment in late stance (Figure 6*b*, My_{late}) were significantly smaller ($p = .029$ and $p < .001$, respectively) for KON compared to NON, whereas the peak flexion moment in early stance (My_{early}) was significantly higher ($p = .004$) for KON versus NON. For the knee twist moment, the internal torsion peak in early stance (Figure 4*f*, Mz_{early}) was smaller for KON in early stance ($p = .007$) and larger for KON in late stance ($p = .003$), compared to NON.

In the tibial frame of reference the same findings were observed for frontal plane and sagittal plane moments. The 2nd peak in adduction moment (Figure 7*a*, Mx'_{late}) and peak extension moment in late stance (Figure 7*b*, My'_{late}) were significantly smaller

Table 5
Comparison of peaks of interest (PoI) during walking with (KON) and without (NON) the dermoskeleton.

PoI†	NON		KON		KON-NON		Sig.	95%CI	
	Mean	SD	Mean	SD	Mean	SD		Lower	Upper
Lab frame of reference									
Mz_{early}	0.46	0.20	0.28	0.20	−0.18	0.11	<0.001*	−0.24	−0.11
Mz_{late}	0.33	0.20	0.49	0.21	0.16	0.11	<0.001*	0.09	0.23
Mz_{term}	−0.41	0.30	−0.19	0.22	0.22	0.17	0.001*	0.12	0.32
My_{early}	4.88	1.64	5.56	1.39	0.68	0.70	0.004*	0.26	1.10
My_{late}	−2.02	1.06	−1.45	1.12	0.56	0.43	<0.001*	0.30	0.82
My_{term}	1.32	0.61	1.50	0.84	0.18	0.51	0.225	−0.13	0.49
Mx_{early}	3.56	0.85	3.45	0.77	−0.11	0.34	0.272	−0.32	0.10
Mx_{mid}	1.16	0.44	1.17	0.56	0.01	0.26	0.857	−0.14	0.17
Mx_{late}	2.37	0.80	2.14	0.81	−0.23	0.34	0.029*	−0.44	−0.03
Tibia frame of reference									
Mz'_{early}	−0.19	0.12	−0.21	0.11	−0.03	0.09	0.324	−0.08	0.03
Mz'_{late}	1.06	0.22	1.13	0.25	0.07	0.22	0.267	−0.06	0.20
My'_{early}	4.69	1.67	5.28	1.49	0.59	0.60	0.004*	0.23	0.95
My'_{late}	−1.86	1.07	−1.21	1.15	0.66	0.48	<0.000*	0.37	0.94
My'_{term}	1.38	0.67	1.49	0.92	0.11	0.46	0.406	−0.17	0.39
Mx'_{early}	3.70	1.23	3.67	1.24	−0.02	0.47	0.861	−0.30	0.26
Mx'_{mid}	1.09	0.37	1.10	0.46	0.02	0.22	0.774	−0.11	0.15
Mx'_{late}	2.25	0.74	1.99	0.73	−0.26	0.33	0.015*	−0.46	−0.06

† M = external knee moment (%BW*HT). Laboratory coordinates: X = anterior–posterior, Y = medio-lateral, Z = vertical. Tibia coordinates: X′ = anterior–posterior, Y′ = medio-lateral, Z′ = superior–inferior.
* Significant at $p < .05$.

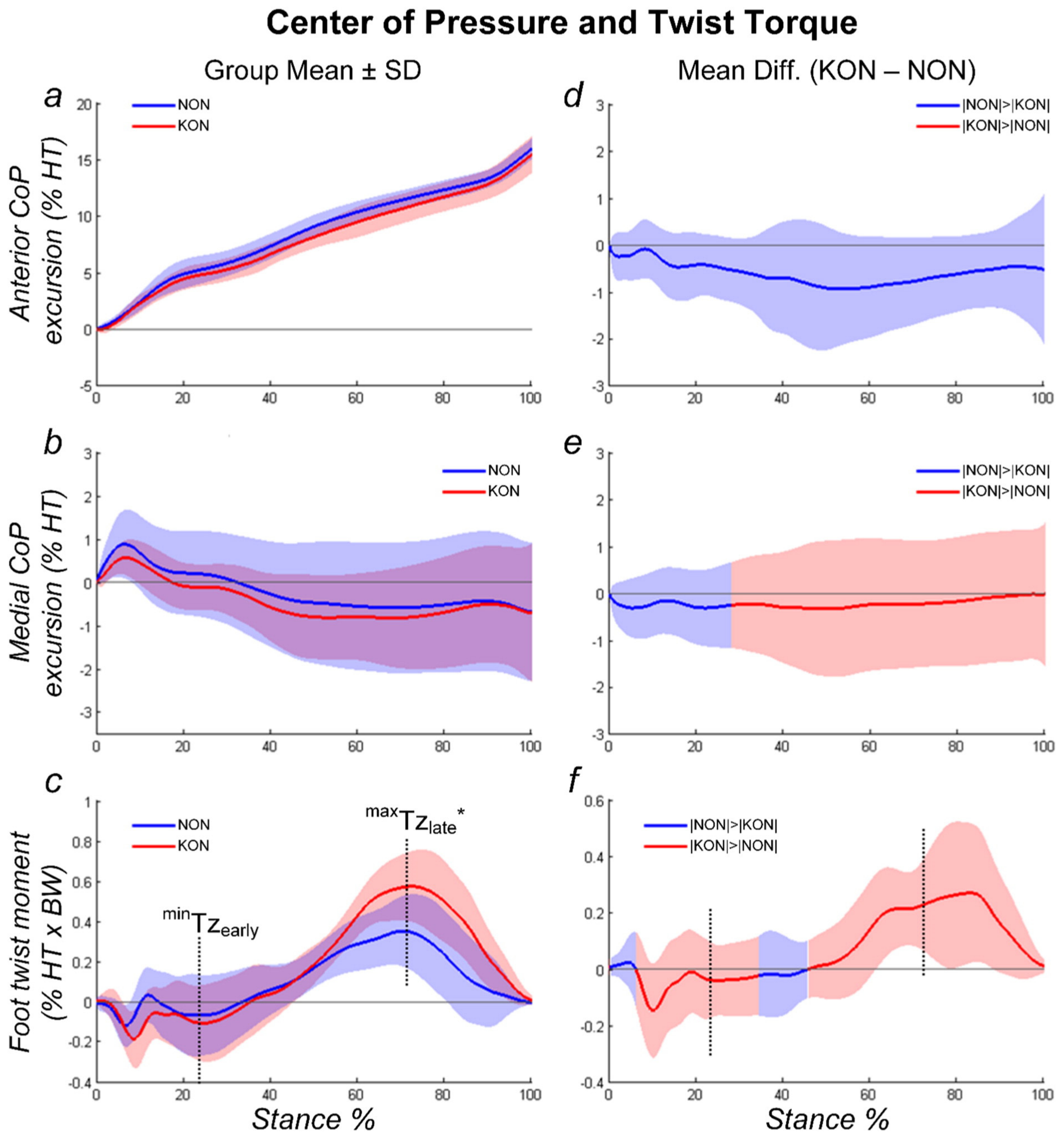

Figure 3. Foot center of pressure (CoP) excursion and ground reaction vertical torque (Tz) for NON (blue) and KON (red) conditions. Column 1 (*a*, *b*, *c*): Solid lines are means across subjects for each condition and shaded regions represent ±1 standard deviation (SD) from the respective means. Column 2: (*d*, *e*, *f*): Solid line represents the mean of the difference between conditions and shaded region represents ±1 SD from the mean of the difference. Color shading indicates if |KON| > |NON| (red) or |NON| > |KON| (blue). Vertical dotted lines and corresponding labels are the parameters of interest (PoI). PoI labels with * indicate a significant difference between conditions ($p < .05$). Row 1: (*a*, *d*) CoP anterior–posterior X excursion; Row 2: (*b*, *e*) CoP medio-lateral Y excursion; Row 3: (*c*, *f*). Ground reaction torque Tz.

($p = .015$ and $p < .001$, respectively) for KON compared to NON, and the peak flexion moment in early stance (My'_{early}) was significantly higher ($p = .004$) for KON versus NON. A major difference, however, was found in the transverse plane moment compared to the moment about the lab vertical axis. In the tibia frame of reference the transverse (internal/external rotation) knee moment was the same for both KON and NON conditions, with no statistically significant differences in their peaks.

Difference plots in Figures 6*d*–*f* and 7*d*–*f* show that the magnitude of the difference in lab referenced knee moments was approximately .5 to one percent BW∗HT in the frontal plane, sagittal plane, and transverse plane, whereas in tibia lab coordinates the transverse plane difference decreases to less than .1% BW∗HT and remains largely unchanged for sagittal and frontal plane moment differences.

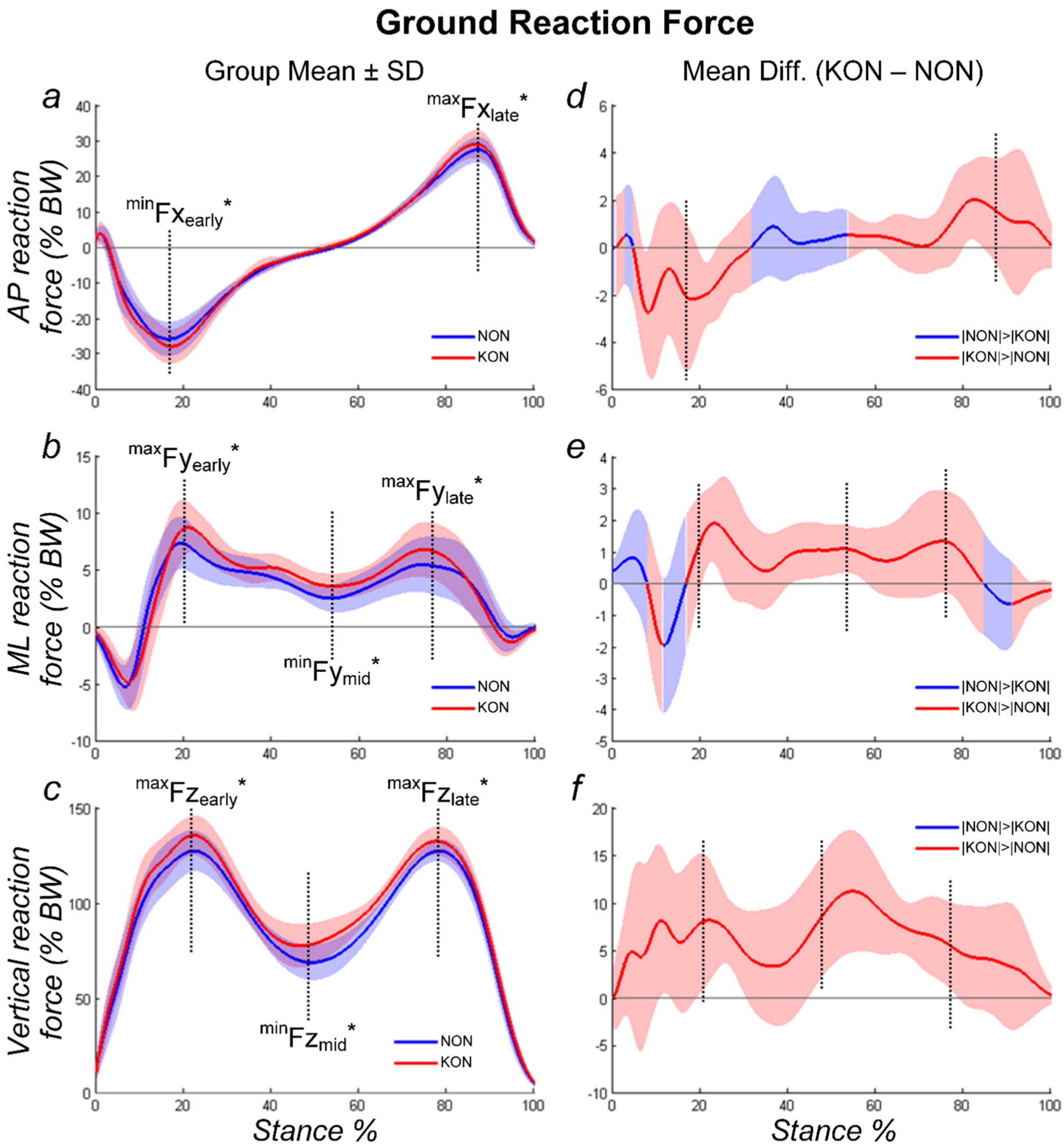

Figure 4. Ground reaction forces (F) for NON (blue) and KON (red) conditions. Column 1 (*a*, *b*, *c*): Solid lines are means across subjects for each condition and shaded regions represent ±1 standard deviation (SD) from the respective means. Column 2: (*d*, *e*, *f*): Solid line represents the mean of the difference between conditions and shaded region represents ±1 SD from the mean of the difference. Color shading indicates if |KON| > |NON| (red) or |NON| > |KON| (blue). Vertical dotted lines and corresponding labels are the parameters of interest (PoI). PoI labels with * indicate a significant difference between conditions ($p < .05$). Row 1: (*a*, *d*) Anterior–posterior ground reaction force, Fx; Row 2: (*b*, *e*) Medio-lateral ground reaction force, Fy; Row 3: (*c*, *f*) Vertical ground reaction force Fz. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

4. Discussion

The objective of this study was to quantify the effects of a lower-extremity dermoskeleton on the biomechanics of the knee during over-ground walking in healthy adults within a controlled laboratory setting. The purpose was to identify the biomechanical mechanisms that alter knee joint kinetics when walking with a dermoskeleton, in order to evaluate if the resulting knee moments during stance phase of gait would be *biomechanically* indicated, or contraindicated, for the KOA patient. The results suggest

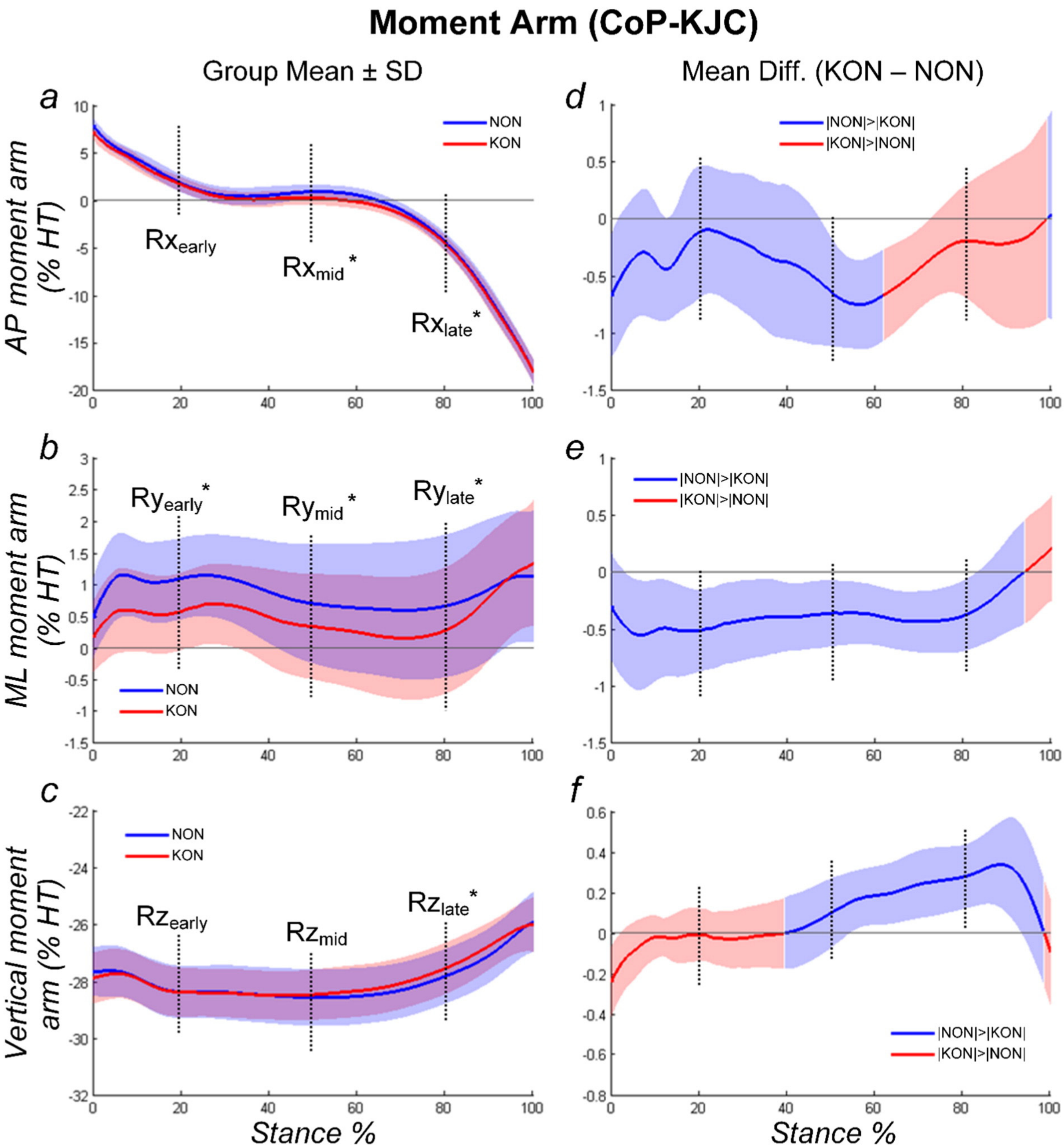

Figure 5. CoP-to-KJC moment arms (R) for NON (blue) and KON (red) conditions. Column 1 (*a*, *b*, *c*): Solid lines are means across subjects for each condition and shaded regions represent ±1 standard deviation (SD) from the respective means. Column 2: (*d*, *e*, *f*): Solid line represents the mean of the difference between conditions and shaded region represents ±1 SD from the mean of the difference. Color shading indicates if |KON| > |NON| (red) or |NON| > |KON| (blue). Vertical dotted lines and corresponding labels are the parameters of interest (PoI). PoI labels with * indicate a significant difference between conditions ($p < .05$). Row 1: (*a*, *d*) Anterior–posterior external knee moment arm, Rx; Row 2: (*b*, *e*) Medio-lateral external knee moment arm, Ry; Row 3: (*c*, *f*) Vertical external moment arm, Rz.

a favorable biomechanical environment for the KOA knee due to the decrease in knee abduction/adduction moment. Whether the device is *clinically* indicated is outside the scope of the current study.

4.1. Is the dermoskeleton biomechanically indicated for KOA?

We found that even though the body stature normalized ground reaction forces and twist moment were greater when wearing the dermoskeleton (due to the added mass when wearing the device), the knee joint moments were regulated by subtle

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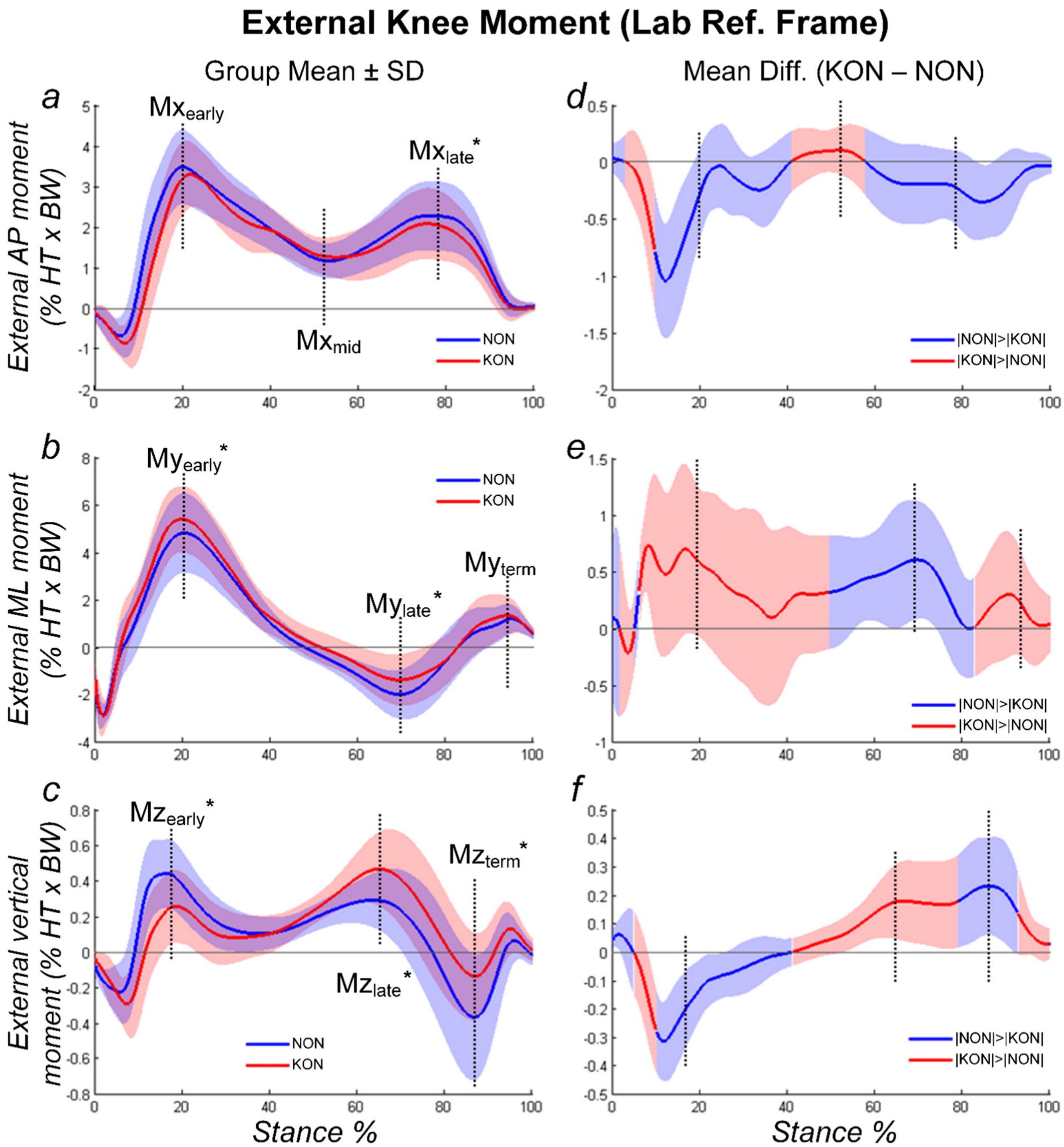

Figure 6. External knee moment (M) in the laboratory frame of reference for NON (blue) and KON (red) conditions. Column 1 (*a*, *b*, *c*): Solid lines are means across subjects for each condition and shaded regions represent ±1 standard deviation (SD) from the respective means. Column 2: (*d*, *e*, *f*): Solid line represents the mean of the difference between conditions and shaded region represents ±1 SD from the mean of the difference. Color shading indicates if |KON| > |NON| (red) or |NON| > |KON| (blue). Vertical dotted lines and corresponding labels are the parameters of interest (PoI). PoI labels with * indicate a significant difference between conditions ($p < .05$). Row 1: (*a*, *d*) Knee adduction/abduction moment about anterior–posterior lab axis, Mx; Row 2: (*b*, *e*) Knee flexion/extension moment about medio-lateral lab axis, My; Row 3: (*c*, *f*) Knee moment arm about vertical lab axis, Mz.

alterations in joint moment arms that acted to reduce some moments while increasing others. We discuss each of these parameters and their potential indication for the OA knee.

4.1.1. Knee abduction/adduction moment

Most notable among the decreases observed in the KON compared to NON condition was the reduction in knee adduction moment when walking with the dermoskeleton, compared to walking without the device. Although statistically significant only for

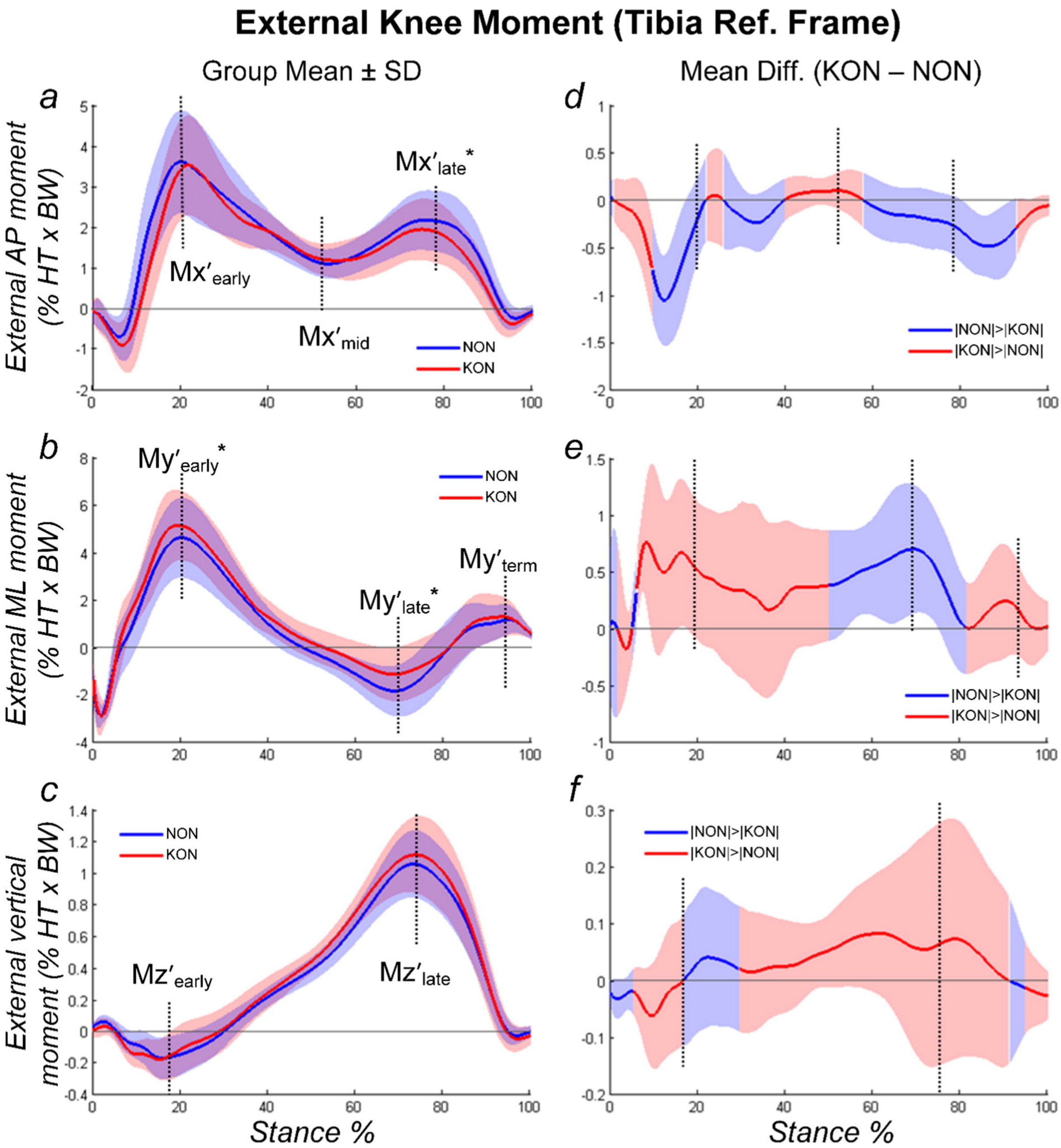

Figure 7. External knee moment (M′) in the tibia frame of reference for NON (blue) and KON (red) conditions. Column 1 (*a*, *b*, *c*): Solid lines are means across subjects for each condition and shaded regions represent ±1 standard deviation (SD) from the respective means. Column 2: (*d*, *e*, *f*): Solid line represents the mean of the difference between conditions and shaded region represents ±1 SD from the mean of the difference. Color shading indicates if |KON| > |NON| (red) or |NON| > |KON| (blue). Vertical dotted lines and corresponding labels are the parameters of interest (PoI). PoI labels with * indicate a significant difference between conditions ($p < .05$). Row 1: (*a*, *d*) Knee adduction/abduction moment about coronal plane axis of knee, Mx'; Row 2: (*b*, *e*) Knee flexion/extension moment about sagittal plane axis of the knee, My'; Row 3: (*c*, *f*) Knee moment arm about transverse plane axis of the knee, Mz′.

the 2nd peak, the difference was maintained regardless of whether the moment was calculated in the laboratory or tibia frame of reference (Figures 6*a* and 7*a*). Knee adduction moment data (in the lab frame) for a single participant are shown in Figure 8*a* (NON condition) and Figure 8*d* (KON condition) where both 1st and 2nd peaks were reduced in the KON condition.

The reduction in the medio-lateral moment arm (Figure 5*b*) is largely responsible for the lower peak KAM. Figure 8 shows that the external moment component Ry∗Fz (medio-lateral moment arm by vertical ground force, thin solid line) is the primary cause of the reduced moment, and in fact compensates for the fact that the other moment component −Rz∗Fy (vertical moment arm by

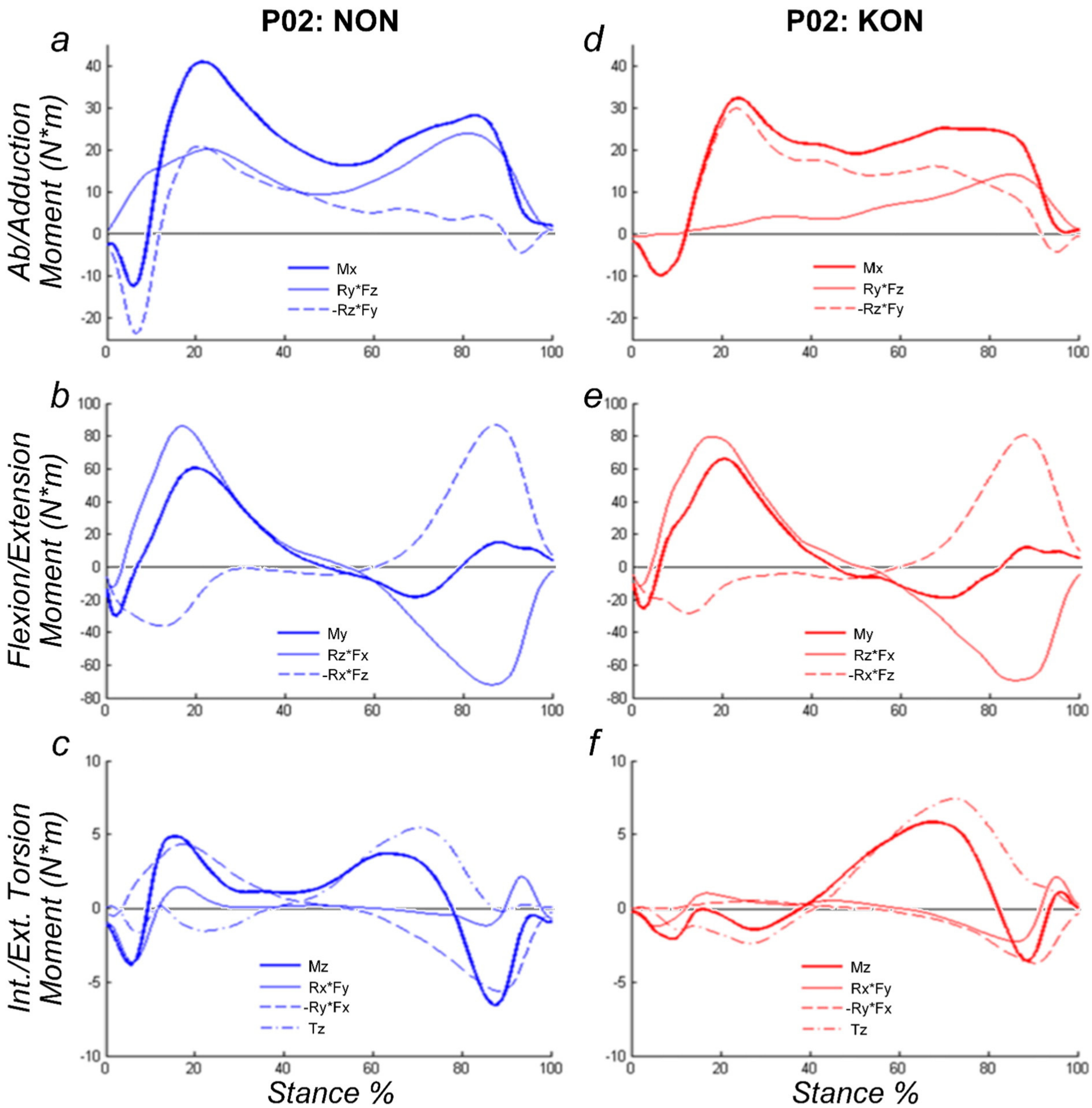

Figure 8. External knee moments and their composition in the laboratory frame of reference for the NON (blue, Column 1) and KON (red, Column 2) conditions for a representative participant. The heavy solid line is the total moment, and the thin solid line and dashed line represent the cross-product moment components that sum to arrive at the total moment, where R refers to moment arm, F refers to ground force, and T refers to twist torque. Row 1: Knee adduction/abduction moment; Row 2: Knee flexion/extension moment; Row 3: Knee vertical axis moment. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

medio-lateral ground reaction force, dashed line) is increased due to the increased Fy magnitude and the fact that Rz (a function of knee height) is relatively invariant (except in terminal stance, Figure 5*c*).

The source of the structural compensation appears to be lateral postural adjustment of the body center of mass as indicated by the increased lateral excursion of the CoP in KON compared to NON condition (Figure 4*b*), suggesting a more laterally positioned center of mass during stance. This compensation may be related to the moderate stiffness of the dermoskeleton hip joints in the coronal plane, but may also be due to participants widening their base of support to increase stability. Indeed, we found that participants' increased their step width by on average 2 cm when using the dermoskeleton (Table 2). Further testing would be required to determine if repeated sessions of walking change these compensations.

What is clear from this study is that the KAM peaks are modified favorably (attenuated) when wearing the dermoskeleton, by creating a favorable biomechanical environment that appears to be linked to lateral CoP excursion [33]. Reduction of the KAM has become a central focus of conservative biomechanical treatment for KOA because of its suspected role in disease progression and symptomology [34–36]. Studies are now needed to determine if the KAM is reduced for symptomatic medial-compartment KOA patients when wearing the dermoskeleton. Studies are also needed to explore how loads are transmitted from the device to the soft-tissues of the human joint to determine if long term use that overly protects the medial compartment could jeopardize the lateral compartment.

4.1.2. Knee flexion/extension moment

The patterns of sagittal plane moment (Figures 6*a* and 7*a*) indicated that the external knee flexion moment in early stance was higher for the KON compared to NON condition, and the opposite was true for the knee extension moment in late stance. The representative data in Figure 8*b* (NON condition) and Figure 8*e* (KON condition) suggest that the $-Rx*Fz$ moment component is smaller, and when summed with the less variable $Rz*Fx$ component, results in a higher moment in early stance and lower moment in late stance. Inspection of the anterior–posterior CoP excursion data in Figure 3*a* suggests that the CoP_x for KON lags slightly in "stance phase time" relative to NON, which in turn delays the zero crossing of the flexion/extension knee moment, increasing the flexion impulse and decreasing the extension impulse. Although not quantitatively analyzed, the deviation in the CoP_x excursion is consistent with a stiffer knee at load acceptance (i.e. the motor resisting flexion).

This could have benefits for neuromuscular training in KOA patients aimed at reducing muscle co-contraction [23,37]. For example, by artificially supplying the additional stiffness that the central nervous system believes the "knee complex" requires for early stance phase stability, the amount of co-contraction patients use to stabilize their joint might be reduced, lowering the joint articular forces during locomotion, and reducing the mechanisms that cause pain in the OA knee. Studies are therefore required to determine if neuromuscular mal-adaptations, such as excessive and/or unbalanced (medial/lateral) co-contraction can be reduced in symptomatic KOA patients whist exercising with a dermoskeleton.

The data from our healthy controls suggest that the modifications in knee flexion/extension moments are potentially favorable for the KOA patient. Several studies of the biomechanics of the OA knee observe a reduced 1st knee flexion moment in early stance [38–41] and in some studies an increased 2nd flexion moment in late stance [41] or an increased extension moment in late mid-stance [42]. The only sagittal plane moment to increase when using the dermoskeleton was the first flexion moment peak. Although it may be argued that increasing the flexion moment in early stance will increase the axial load of the femoro-tibial joint, it is likely this increase is significantly smaller than the considerable co-contractive forces that can be generated in the knee and which cannot be observed in the net moments from inverse dynamic analysis. As described above, data from this study may be useful for controlling the relative flexion/extension impulses to assist the patient in regulating agonist–antagonist activity. Studies are needed to test these ideas.

4.1.3. Knee internal/external (torsion) moment

Unlike frontal and sagittal plane moments, the patterns of knee torsion moment (Figures 6c and 7c) differed considerably according to whether it was expressed in the laboratory or anatomical reference frame, consistent with other studies [43,44]. Although the findings from the laboratory frame of reference suggest considerable differences between the KON and NON conditions, where knee torsion moments were higher for KON throughout stance phase, when transformed into tibia frame of reference (and ultimately more relevant to knee function) they were found to be equivalent. This suggests that reference frame selection is indeed an important factor in quantifying the effects of orthotic technologies, whether passive or active, on the biomechanics of the knee.

4.2. What might be the benefits of the dermoskeleton for a clinical population?

Conservative management of knee osteoarthritis (KOA) with both pharmacologic (analgesics, anti-inflammatories, corticosteroid injections etc.) and non-pharmacologic treatments (weight loss, exercise, bracing, footwear, etc.) can reduce symptoms of pain [45,46], but approximately one in four of these patients will not respond to conservative management [14]. These KOA patients typically live with their condition for nine to 12 years before opting for total knee arthroplasty (TKA) [21,47]; the so-called "treatment gap" [48] is estimated to be even longer than this (>20 years) for younger KOA patients [14]. Continued exercise and physical activity is critical for reducing symptoms of pain and maintaining health and physical function in these patients [49–51], but there are many challenges to accomplishing this in the KOA population [52–54]. Our results suggest that dermoskeletons, which are designed to reduce knee loading and enable greater levels of physical activity, could address this treatment gap.

A significant factor that limits the success of exercise and physical activity-based therapies is that the biomechanical environment of the pathological joint contraindicates loading the joint [55]. Frontal plane misalignment, and the resulting increase of the knee adduction moment (KAM), is a strong predictor of progression of medial compartment KOA [36,56], is non-responsive to muscle strengthening [57,58], and is positively associated with future chronic knee pain in older adults [59]. Structural reduction of the KAM has thus become a focus of conservative KOA treatment, typically with orthotic devices such as bracing [60–63] or lateral wedge footwear or insoles [62,64–69]. Although bracing and wedged insole solutions show promise for reducing the KAM, it is unknown if these passive solutions can address the neuromuscular deficits acquired by many KOA patients in the decade or two prior to electing TKA. Neuromuscular impairments that arise from KOA are also linked to pain [37] and may play a role in disease progression [23] and worse outcomes following TKA [26,27].

Unfortunately there are few options for delivering novel sensorimotor therapies for KOA patients that are also able to dynamically stabilize the knee during locomotion. Although over-ground robotic exoskeletons have been commercially available for almost a decade [70] and have been researched extensively for neuro-rehabilitation of stroke and spinal cord injury [71–73], they have not yet been considered in the treatment of KOA or other musculoskeletal disorders. The potential benefit of over-ground robotic exoskeleton technology, however, could be significant. The ability to exercise safely could improve physical activity levels and core muscle strength, and also opens the door to new interventions for improving sensorimotor deficits in people with musculoskeletal impairment.

4.3. Limitations

There are many limitations to this study. One is that the sample was small (n = 13), young (27 ± 6 years old), and all participants were asymptomatic. Therefore the findings may not be generalizable to an older, gait impaired population. Nevertheless, the fact that the dermoskeleton demonstrated a biomechanical environment that does not appear to be contraindicated for KOA allows us to conclude that biomechanical study of a clinical population is warranted.

Gait analysis is fraught with potential sources of error, not all of which can be completely controlled. Although we tracked body segments with marker cluster arrays that were maintained in place in both the KON and NON conditions (to eliminate variability caused by moving the markers), it is possible that the static orientation of clusters of skin markers on the shank and thigh were impacted by the presence of shin cuffs in the KON condition, and introduced a systematic bias. The careful static calibration procedure we used [72] should have eliminated any such biases, but cannot account for dynamic effects that may arise due to differences in local tissue stiffness when straps and braces are adjacent to the marker clusters.

Related to this, and limitations in marker placement due to the presence of the dermoskeleton suspension and bracing system, several commonly used anatomical landmarks that are important for model generation (such as the ASIS of the pelvis, lateral epicondyles of the knee, greater trochanter, etc.) could not be tracked directly. A previous study from our lab showed the sacral plate is a viable alternative when conventional ASIS tracking is not possible [27], and using static trials and virtual registration of anatomical landmarks to limb clusters is an accepted approach to overcoming this limitation and has been used for decades [73].

In this study, our simplified knee-wrench analysis did not include the contributions of the interaction force and moments between the human and the dermoskeleton. The dermoskeleton tested in this study is designed with a frontal-plane pin joint at the knee which cannot support a moment, and soft tissue motion between the device and the skeletal structure likely precludes the transmission of significant transverse-plane loads. Future studies are required to analyze the role of passive, and active, dermoskeleton loads on the net knee joint moments. Additionally, although the net external joint moments are correlated with internal joint loading [74,75], they do not capture much of the variance due to compressive action of muscle forces. As the dermoskeleton technology poses a unique potential to modify neuromuscular control strategies, future work and detailed musculoskeletal models of the knee [76–78] must be used to fully characterize changes in joint loading due to powered exoskeleton assistance.

5. Conclusions

We conclude that the external moment patterns experienced by the healthy knee when walking with a dermoskeleton indicate a biomechanical environment that can reduce or limit the knee adduction moment, with minimal disruption to transverse plane moments. The mechanism explaining this reduction was a lateral shift in the center of pressure that reduces the knee adduction moment arm. Beyond the expected increase in early stance loading due to increased inertia, kinetics in the sagittal plane suggest the dermoskeleton may also be useful for modifying neuromuscular control by regulating flexion/extension impulses without increasing the 2nd flexion moment and potentially decreasing the late mid-stance knee extension moment. Further study of this technology is warranted to determine its potential for delivering biomechanical and neuromuscular gait interventions for patients with KOA.

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Conflict of interest

No part of the present work has been published elsewhere and there is no commercial interest of any authors related to this work.

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