Patients with medial knee osteoarthritis reduce medial knee contact forces by altering trunk kinematics, progression speed, and stepping strategy during stair negotiation: A Pilot Study

Journal:	*Journal of Applied Biomechanics*
Manuscript ID	JAB.2017-0159.R3
Manuscript Type:	Original Research
Keywords:	clinical biomechanics, cartilage, motion analysis, knee osteoarthritis, musculoskeletal modeling, knee contact forces

Title: Patients with medial knee osteoarthritis reduce medial knee contact forces by altering trunk kinematics, progression speed, and stepping strategy during stair ascent and descent: A Pilot Study

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Conflict of Interest Disclosure: None.

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Abstract

Medial knee loading during stair negotiation in individuals with medial knee osteoarthritis, has only been reported in terms of joint moments, which may underestimate the knee loading. This study assessed knee contact forces (KCF) and contact pressures during different stair negotiation strategies. Motion analysis was performed in five individuals with medial knee osteoarthritis (52.8±11.0 years) and eight healthy subjects (51.0±13.4 years) while ascending and descending a staircase. KCF and contact pressures were calculated using a multi-body knee model while performing step-over-step at controlled and self-selected speed, and step-by-step strategies. At controlled speed, individuals with osteoarthritis showed decreased peak KCF during stair ascent but not during stair descent. Osteoarthritis patients showed higher trunk rotations in frontal and sagittal planes than controls. At lower self-selected speed, patients also presented reduced medial KCF during stair descent. While performing step-by-step, medial contact pressures decreased in osteoarthritis patients during stair descent. Osteoarthritis patients reduced their speed and increased trunk flexion and lean angles to reduce KCF during stair ascent. These trunk changes were less safe during stair descent where a reduced speed was more effective. Individuals should be recommended to use step-over-step during stair ascent and step-by-step during stair descent to reduce medial KCF.

Keywords: Knee osteoarthritis, motion analysis, knee contact forces, contact pressures, musculoskeletal modeling.

Word Count: 3996 words.

Introduction

Stair negotiation and level walking are common activities of daily living. However, stair negotiation is biomechanically more challenging [1], demanding a higher range of motion in the lower extremity [2], higher moments acting at the knee joint [3-5] and, consequently, increased quadriceps demands compared to level walking. Thus, stair negotiation is particularly demanding for the elderly or subjects with knee osteoarthritis (OA) [6], who often face the first difficulties in daily task performance and pain complaints [7], particularly during stair descent [8]. However, stair negotiation has not been deeply explored in OA with most studies in literature focusing on knee loading during level walking as a biomarker for OA onset and progression. Previous literature has shown reduced knee flexion moments (KFM) [4,7,9], non-conclusive findings in knee adduction moments (KAM) [4,10] and altered muscle activation patterns [6] in severe knee OA patients during stair negotiation. In addition, these patients exhibited higher trunk flexion angles [11,12] and hip flexion moments [11,13] than healthy subjects while ascending stairs [11]. These alterations observed in OA patients have been associated with a loss of quadriceps function [14,15] as these muscles provide extensor moments necessary to accelerate the upward propulsive phase occurring during the first part of stair ascent and to decelerate the lowering of the body during stair descent [16].

Generally, healthy and young individuals use a traditional step-over-step motion pattern during stair negotiation, but OA patients frequently feel forced to adjust their stair gait due to knee pain, reduced range of motion, muscle weakness, stiffness and instability complaint [17,18]. Therefore, they often adopt alternate walking patterns, such as increased handrail use, sideways motion, or a step-by-step patterns (placing both feet on the same step) [19,20,21] and/or a significantly reduced speed [4,22]. In healthy subjects, the step-by-step strategy has been demonstrated to require higher energy costs and be less efficient than step-over-step, while it seems to increase stability and compensates for lower-limb weakness [19,23]. However, significant reductions in KFM were found for the leading leg during step-by-step when compared to step-over-step while descending stairs in healthy subjects, and reduced anteroposterior force for step-by-step versus step-over-step either during stair ascent or descent [23].

To date it is still unknown how the altered stair negotiation patterns observed in individuals with knee OA affect the compartmental knee contact forces (KCF) as only kinematics and kinetics [4,11,24] have been explored, which do not provide direct measures of cartilage loading. However, KCF reflect not only the influence of external forces but also the muscle and ligament forces. Computational approaches are non-invasive, do not alter the knee biomechanics and can be applied to a larger number of subjects compared to *in vivo* KCF calculations. Therefore, computation of KCF in patients with knee OA during gait has received increasing attention over the last years [25-27]. Previous research [28] has shown the important role of muscle action controlling flexion-extension and adduction-abduction moments in joint loading, specially, during late stance of gait. This was particularly evident in patients with established knee OA. To our knowledge, however, KCF calculated using musculoskeletal modeling that accounts for muscle and ligament forces in combination with dynamic simulations has never been used in individuals with knee OA during stair negotiation. Therefore, the effectiveness of the observed speed reduction [4,22] and changes in stepping strategy in controlling knee joint loading during stair negotiation is unexplored.

The first objective of this study was to compare knee joint loading and trunk kinematics during stair ascent and descent in individuals with medial knee OA against healthy subjects during step-over-step at controlled speed. We hypothesize that OA patients will present lower knee loading than healthy subjects trying to avoid pain. The second objective was to investigate the influence of stair negotiation strategy on knee joint loading magnitude and distribution when individuals performed step-over-step at their preferred speed or were using step-by-step.

We hypothesize that by reducing stair walking speed or by using step-by-step instead of step-over-step, patients will reduce the KCF and redistribute the knee loading to avoid the overloading of the involved compartment.

Methods

Five participants (2 females and 3 males) were recruited for this study via a volunteer database diagnosed in clinical practice with symptomatic bilateral medial knee OA. Eight participants (4 females and 4 males) were recruited on a volunteer basis from the university context, who were asymptomatic and had no history of OA. Participants underwent magnetic resonance imaging (MRI) and completed the Hip [29] and Knee [30] disability and Osteoarthritis Outcome Score questionnaires (Table 1). The Research Ethics committee for Science & Engineering at the Metropolitan Manchester University approved the study. Participants signed the written informed consent form prior to participation.

Patients were classified as having mild (1) moderate (2) and severe (3) knee OA based on pain complaints and three parameters observed on the MRI (Table 2): cartilage defect; bone marrow lesion (BML); and presence of osteophytes. Cartilage was scored for partial and full thickness loss as a % of the surface area in which: 0 when none; 1 when < 15% of cartilage loss; 2 when 15-75% of cartilage loss; 3 when >75% of cartilage loss in a region (medial, lateral or patellofemoral). BML size was scored as follows: 0 when none; 1 when BML size <1 cm; 2 BML when size >1 cm; 3 when multiple BML. Presence of osteophytes was scored based on their size as follow: 0 when no osteophytes; 1 when size < 5mm; 2 when size < 1cm; 3 when > 1cm. All patients presented with bilateral medial knee OA classified as moderate to severe by a consultant radiologist.

Motion analysis was performed while barefoot ascending and descending a staircase consisting of seven 17.2cm-height steps (Figure 1). A 10-camera 3D motion capture system (Vicon Motion Systems Inc, Los Angeles, CA, USA) synchronized with four force platforms (embedded in the middle of the staircase) recorded the 3D position of 34 reflective markers according to an extended lower-body plug-in-gait marker set protocol [31] at 100 Hz, and measured ground reaction forces at 1000 Hz (Kistler, Amherst, New York, United States).

Ground reaction forces were filtered using a second order Butterworth low pass filter, with cut-off level at 30Hz, and marker trajectories using a smoothing spline with cut-off at 6Hz.

Six trials per condition were collected for ascending and descending for step-over-step at controlled speed, *i.e.* alternating feet per step (Figure 1) with cadence controlled by a metronome at 90 beats per minute, corresponding to the normal self-selected stair walking speed in healthy subjects [32]. Furthermore, two alternative strategies were tested: step-over-step at self-selected speed; and step-by-step at self-selected speed, *i.e.* both feet per step (Figure 1). The use of the handrail was not allowed. For safety reasons, patients wore a harness during the data collection.

A multi-body knee model with 6 degrees of freedom for the tibiofemoral and patellofemoral joints and fourteen ligaments was used. <u>More details about the model can be seen in the supplementary material.</u> Development and validation of the knee model are detailed elsewhere [33]. The model included an elastic foundation formulation [34] to compute cartilage contact pressures. This model was integrated into an existing lower extremity musculoskeletal model [35] with 44 musculotendon units.

The lower extremity model was scaled to subject-specific segment lengths as determined in a static calibration trial. The joint angles were computed using an inverse kinematics algorithm. The concurrent optimization of muscle activations and kinematics (COMAK) algorithm [33,36], was used to compute the secondary tibiofemoral and patellofemoral kinematics, muscle and ligament forces, and contact forces by minimizing the muscle volume weighted sum of squared muscle activations plus the net knee contact energy. The COMAK algorithm modulates muscle excitations to track knee flexion, while secondary knee motions (tibiofemoral translations and non-sagittal rotations) and patellofemoral kinematics evolve naturally due to muscle, ligament, cartilage contact, and external loading. The secondary tibiofemoral kinematics and patellofemoral kinematics are load-dependent as they evolve as a function of muscle and ligament forces, and cartilage contact. Tibiofemoral and patellofemoral cartilage contact pressures were computed using an elastic foundation model [34] in which pressure is assumed to be a function of the depth of penetration between contacting cartilage surfaces. Depths of penetration for each triangle in a mesh were determined at each time step using ray-casting techniques [37]. At each triangle of the tibia plateau, the contact pressure was computed, in which cartilage was assumed to have an elastic modulus of 10 MPa, a Poisson's ratio of 0.45, and a uniform thickness of 2 mm for each surface [34]. Subsequently, an inverse dynamics algorithm computed the external joint moments.

The knee model performance has previously been validated. As kinematic validation, the predicted joint kinematics in the secondary degrees of freedom of the knee were validated against joint kinematics measured using dynamic MRI and are reported in the study of Lenhart *et al.*[33]. As dynamic validation, the calculated KCF were compared with instrumented implant data provided through the Grand Challenge Competition to Predict in vivo Knee Loads, a subject-specific data set that allows researchers to validate muscle and contact forces estimated in the knee. When comparing between the measured and calculated KCF, the joint contact load prediction errors (root-mean-square (rms) error = 0.33 BW) [36,38] were comparable to those (rms error = 0.26 BW) observed from a unique optimization approach, termed force-dependent kinematics, introduced by the 2014 "Grand Challenge" winner [39], and slightly better than those that have been obtained using traditional optimization or forward dynamic simulations [40,41].

Calculated KCF were normalized to body weight (BW) and moments to the product of body weight and height (BW×Ht). All data were time normalized to the stance phase (*i.e.* from initial contact to toe off collected from either of the four force plates).

KCF, moments and angles throughout the stance phase were averaged over all trials for each leg. Trunk angles were calculated relative to the ground reference frame. The highest peaks during the first and second half of the stance phase for stair ascent and descent respectively, were determined for the total KCF, medial KCF, and lateral KCF. The highest peak KFM, KAM were determined for all activities whereas peak knee rotation moment (KRM) were only clear for step-over-step tasks while ascending. Furthermore, maximum contact pressures in the medial tibial plateau were assessed at the instant of peak medial KCF.

Independent-samples *t*-test (SPSS Inc., v17.0) evaluated the significance ($p < .050$) of the differences in peaks (tested for normality by Kolmogorov-Smirnov and Shapiro-Wilk) between the two groups and paired-samples *t*-test between strategies (step-over-step at controlled *versus* self-selected speed, and step-over-step *versus* step-by-step) within each group.

As maximum contact pressures did not show a normal distribution, the non-parametric Mann-Whitney-U test was used to evaluate the significance ($p < .050$) of the differences between the two groups. Wilcoxon matched-pair test ($p < .050$) tested the significance of the differences between strategies within each group.

Results

Age, body mass and height, and also speed did not differ significantly between the two subject groups (Table 1). The medial OA group had significantly more knee pain ($p < .001$) and significantly higher function limitations in activities of daily living ($p < .001$) than controls. Level of knee pain was highly correlated with function limitations (R > 0.87).

Peak medial KCF (1.86±0.54, $p < .000$) and lateral KCF (1.52±0.36, $p = .015$) were significantly lower in individuals with OA compared to controls (2.51±0.28 and 2.24±0.81, respectively, medial and lateral KCF) during stair ascent (Figure 2). During stair descent, on the other hand, no significant differences were observed between the two groups (S1 Figure and S1 Table). Maximum contact pressures were also lower in individuals with OA, during stair ascent, however, not statistically significant (Table 4). Patients with OA exhibited more similar pressures during stair ascent and stair descent, whereas control subjects clearly reduced pressures from ascending to descending (Figure 4).

Individuals with OA exhibited significantly lower peak KFM during both stair ascent (0.050±0.017, $p = .002$) and descent (0.058±0.018, $p = .022$) compared to controls (0.070±0.012 and 0.073±0.015, respectively, at stair ascent and descent). No significant differences in the peak KAM or KRM were observed between the two groups (S2 Figure and S2 Table).

Individuals with OA had higher trunk flexion angles (24.45±3.76, $p = 0.001$ during stair ascent) and tended to lean the trunk more towards the leading leg in the frontal plane (2.76±1.38, $p = 0.069$ during stair ascent) throughout the stance phase compared to controls (18.43 (3.74) and 0.93 (2.82), respectively, trunk flexion and trunk lean angles during stair ascent) during both stair ascent and descent (Figure 3 and S3 Table). During stair descent, the OA group exhibited a larger variation between subjects in the trunk kinematics in the frontal and transversal planes, shown by the high standard deviations, compared to controls. In all planes of motion, kinematics of the hip, knee and ankle joints showed a similar pattern of movement between the two groups during stair ascent and descent (S3 Figure and S4 Figure).

When changing speed, all subjects walked slower when they could walk at their preferred speed in comparison with the controlled condition, however only significantly during stair ascent. During stair ascent at decreased speed, the peak medial KCF (p = .024) and lateral KCF decreased (p = .002) in individuals with OA (S6 Figure and S4 Table), whereas the opposite was found for peak lateral KCF (p = .009) in healthy subjects (S5 Figure and S4 Table). During stair descent, no significant differences in KCF were observed between step-over-step at controlled and self-selected speed in healthy or OA groups. No differences were observed in terms of maximum contact pressures between step-over-step at controlled and self-selected speed in both groups (Table 4).

With reduced speed, patients with OA maintained the increased trunk flexion and lean angles towards the leading leg during stair ascent. During stair descent, on the other hand, OA patients exhibited a smaller variation in the trunk kinematics in the frontal and transversal planes as the speed decreased (S5 Table).

When changing strategies from step-over-step to step-by-step, both controls and OA significantly reduced the speed while ascending (p < .001 and p = .009, respectively) and descending stairs (p < .001 and p = .008, respectively) (Table 3). Both controls (p = .016) and OA (p = .040) exhibited significantly higher peak lateral KCF when using step-by-step instead of step-over-step during stair descent. During stair ascent, however, individuals with knee OA significantly increased the peak medial KCF (p = .008) when using step-by-step, whereas no significant differences were seen in controls (S4 Table). By altering from step-over-step to step-by-step, maximum contact pressures were not significantly different neither in controls or patients with OA (Figure 4 and Table 4) during stair ascent. However, during stair descent maximum contact pressures significantly decreased in patients with OA when using step-by-step (p = .007).

Discussion

This study investigated knee joint loading in terms of magnitude of KCF and cartilage pressures, during stair ascent and descent in individuals with medial knee OA. Using a multibody musculoskeletal model, we showed that patients with OA exhibited reduced tibiofemoral loading during stair ascent, but not stair descent. The reduced contact force during ascent was achieved by increasing the trunk flexion angle, which reduced the knee flexion moment and thus muscle forces compressing the joint. This strategy was not as effective in stair descent, where the trunk was more vertical, thus the knee flexion moment cannot be modulated without large adjustments to trunk flexion that compromise stability on stairs. Furthermore, different strategies in stair negotiation, such as reduction in speed, and employing step-by-step instead of step-over-step were shown to be effective in reducing the knee contact loading.

Our results confirmed the hypothesis that OA patients would present lower KCF than controls. During stair ascent, when asked to walk at controlled speed, which was significantly higher than their preferred speed, the OA group could effectively reduce both peak medial and lateral KCF compared to control subjects. This was possible by executing higher trunk flexion and higher trunk lean towards the leading leg compared to controls. By positioning the centre of mass further forwards and more towards the leading leg at a time where knee is considerably flexed, which potentially induces elevated joint moments, OA patients direct the ground reaction force vector closer to the knee joint centre and, therefore, they reduce the KFM (significantly) and KAM. In addition, the increased trunk flexion decreases the demand on the knee extensors, which generate the propulsion required during stair ascent. Previous studies have also found reduced KFM [4,11] and increased trunk flexion [11] during stair ascent [4,11] and descent [4] in OA patients compared to controls. Despite the reduced KCF, OA patients still reported significantly higher pain complaints compared to controls. Our study is therefore the first to determine that the altered stair walking pattern used by patients with OA, more specific the higher trunk flexion and reduced KFM, effectively unloads the knee joint as reflected in the reduced compartmental KCF.

During stair descent, the compensatory mechanisms used by the OA group were less effective in reducing the knee loading than during stair ascent, and reductions in peak medial and lateral KCF were not statistically significant. Patients could not increase the trunk flexion angles during stair descent as much as they did during stair ascent compared to a healthy control, probably due to fear of falling. During stair descent, the body has to adopt to a more upright position to maintain balance and, therefore, by leaning the trunk too far forwards, patients could compromise their balance [42] and, ultimately fall. The inability to reduce KCF during descent may explain why patients experience higher knee pain [43] during stair descent than ascent.

The second hypothesis that OA patients would be able to reduce the KCF by reducing the speed or by using step-by-step instead of step-over-step strategy has been partially confirmed. When subjects walked at their preferred speed, which was significantly slower than that at controlled execution during stair ascent, individuals with OA significantly reduced medial KCF compared to those occurring at controlled speed, whereas controls kept similar KCF at the medial compartment. When forced to increase their speed, some OA patients felt forced to lean and rotate their trunk more, resulting in a high variation between subjects in the trunk kinematics in these two planes during stair descent. This shows that some patients felt forced to use another mechanism rather than increased trunk flexion to perform stair descent when speed was enforced. This suggests that it is more effective for patients to reduce medial compartment loading during stair descent by reducing the walking speed than by altering trunk kinematics. During stair ascent, on the other hand, the changes in the trunk kinematics were still effective for OA patients to reduce knee loading, even at a higher stair walking speed. In addition, speed reduction allowed OA patients to decrease maximum medial compartment contact pressure. Thus, a reduction in speed together with changes in trunk kinematics are the key strategies used to reduce the knee loading during stair ascent, and a reduction in speed is even more important to reduce the medial knee loading during stair descent.

By changing the stepping strategy and performing step-by-step instead of step-over-step during stair ascent, OA patients significantly increased the medial KCF, even at significantly lower speeds. This resulted from the fact that performing step-by-step, body tends to adopt a straighter position. On the other hand, during stair descent, by performing step-by-step instead of step-over-step, they significantly decreased the medial knee contact pressures. Similarly, Reid *et al.* [23] reported that in healthy subjects, step-by-step strategy was more efficient in reducing the peak KFM when compared to step-over-step strategy during stair descent than stair ascent. From our findings, it is suggested that, in OA patients, step-by-step is only effective in reducing the medial knee loading during stair descent, but not during stair ascent.

The magnitude of KCF in healthy subjects in the present study was higher for stair ascent than those from literature based on measured KCF in subjects with instrumented prosthesis [44,45]. Our controls exhibited an averaged peak total KCF of 4.41 (0.78) BW and 4.20 (0.74) BW for, respectively, stair ascent and descent, whereas Kutzner *et al.* [44] reported averages of 3.16 BW and 3.46 BW for the peak resultant force. Similar results, ranged from 2.90 to 3.50 BW, were reported by Heinlein *et al.* [45]. More similarly, our OA group exhibited peak KCF of 2.78 (0.62) BW and 3.29 (1.14) BW for stair ascent and descent, respectively. Previous simulation studies on healthy subjects and those having TKR during stair ascent, presented compressive joint reaction forces up to 4.00 BW [46]. Differences might be due to several reasons: instrumented implant studies report on patients having TKR and an altered gait pattern may therefore be present; none of the mentioned studies report stair walking speed nor the step height.

The findings of this study should be viewed in light of the following specific limitations. We used a single generic knee model that was scaled to represent the anthropometry of the subjects instead of considering the subject-specific articular geometries, including those of the tibia plateau. Subject-specific articular geometries, muscle-tendon and ligaments properties were not considered in our approach since there was no data available for the cohort studied. Therefore, our model does not account for OA induced changes in the articular geometry, thickness and mechanical properties of the cartilage or changes in the ligaments. Consequently, the reported differences in KCF and contact pressures only result from altered kinematic and kinetic behavior. Bone deformities, ligament laxity or changes in cartilage induced by joint degeneration were not taken into account although they may affect the calculated contact pressures. However, the effect of having a 2-mm constant cartilage thickness instead of a variable thickness on tibiofemoral contact pressure during gait has been previously assessed and showed limited effect on the observed peak contact pressure (about 4%) [47]. Furthermore, although the validation of the model has shown a good agreement between the calculated and experimental kinematics and contact forces in healthy subjects and patients following total knee replacement [33], this validation cannot easily be extended to an OA population. Therefore, this model might present specific limitations when used in patients with knee OA, especially those known to present increased co-contraction (Kellgren-Lawrence score ≥2) resulting in an underestimation of the joint loading [48]. In this model, the ligaments are represented as nonlinear spring elements, one-dimensional discrete elements, rather than deformable 3D representations that account for spatial variations in strain. Instead, some wrapping surfaces were included to improve wrapping around the bony structures but no ligament–ligament interactions were incorporated. The thickness of the cartilage surface was assumed constant, which is a simplification since cartilage thickness varies. This simplification might result in differences in terms of contact pressures and contact areas [49]. Further, the knee model does not include menisci, which are known to distribute pressure in the tibiofemoral joint. Therefore, the absence of menisci might increase the peak contact pressures in the knee joint surface. Finally, it is known that the definition of other lower limb joints influences knee kinematics as well [50], especially the ankle joint, for which only one degree of freedom was considered.

In conclusion, during stair ascent, OA patients could effectively reduce the knee joint loading by reducing their speed, increasing the trunk flexion and lean the trunk more towards the leading leg. However, during stair descent, changes in the trunk flexion and frontal lean were more limited and less effective, requiring reduced speed or even more increased trunk rotation and lean to effectively reduce the peak medial KCF and the contact pressures on the tibia plateau. Furthermore, this study suggests that, in OA patients, step-over-step is more effective in reducing the medial knee loading, particularly at reduced speed, during stair ascent, while step-by-step is more effective during stair descent. Understanding how these compensatory mechanisms work across the whole body can help underpin recommendations on alternative strategies for avoiding overloading of other joints.

Author contributions

All authors take responsibility for the integrity of the work as a whole, including data and accuracy of the analysis. Design: S. Meireles, N. Reeves and I. Jonkers. Conception: S. Meireles, N.D. Reeves, R.K. Jones, C.R. Smith, and D.G. Thelen. All the authors contributed to the analysis and interpretation of the data, drafting of the article and final approval of the article.

Acknowledgements

Integrated in Move-Age program funded by the European Commission as part of the Erasmus Mundus program. The authors thank Dr. Richard Hodgson, PhD from The University of Manchester, UK for his collaboration in this study.

Competing interests

All authors declare no conflict of interest.

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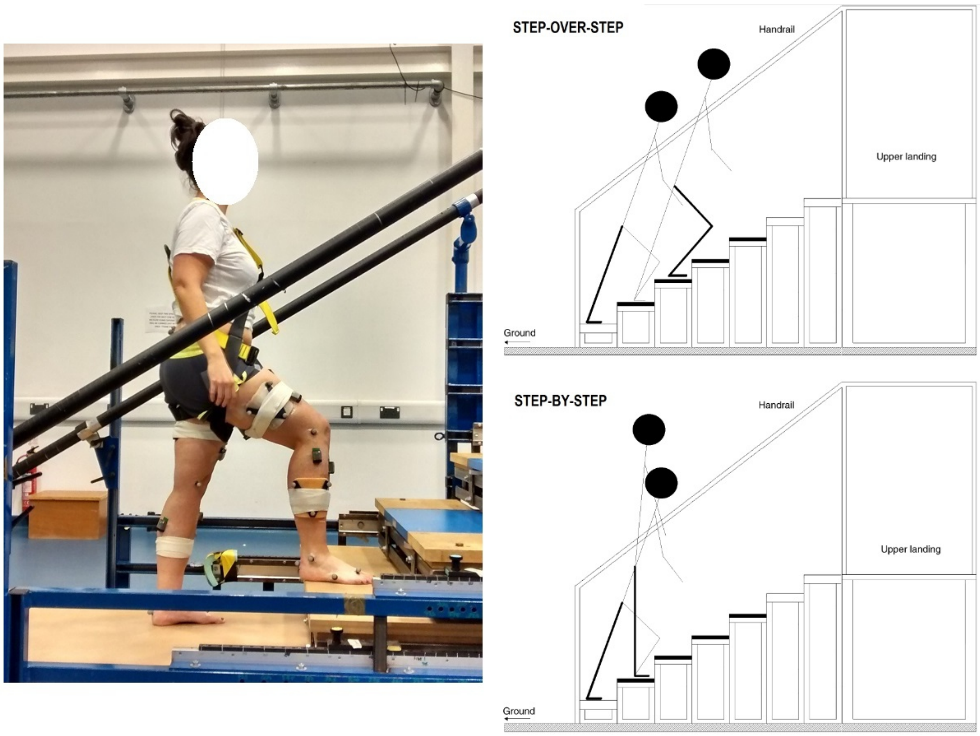

Figure 1 - Marker set on a representative subject while ascending the staircase (left) and a representative scheme of the step-over step (above right) and step-by-step (below right) tasks.

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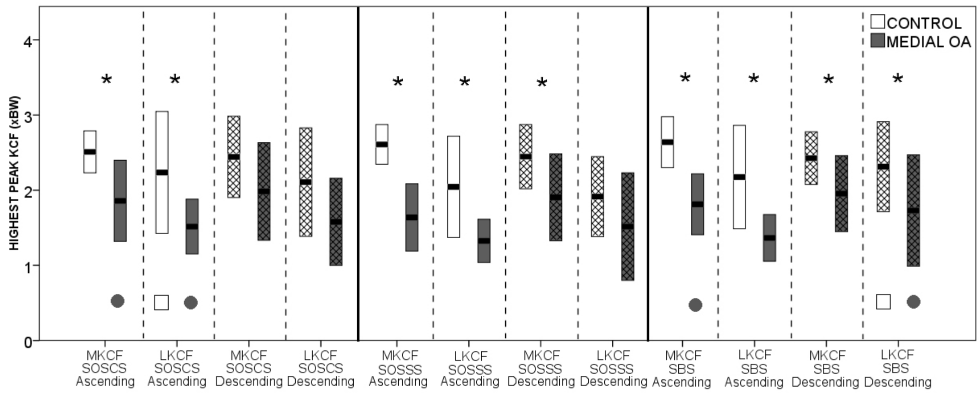

Figure 2 - Peak medial KCF (MKCF) and lateral KCF (LKCF), comparing the two groups of subjects while performing different tasks: step-over-step at controlled speed (SOS CS), step-over-step at self-speed (SOS SS) and step-by-step (SBS) while ascending or descending stairs. * indicates a significant difference between the groups. □ indicates a significant difference between the task in which there is this indication and the task step-over-step while ascending stairs for the control group, whereas ● is used to the OA group.

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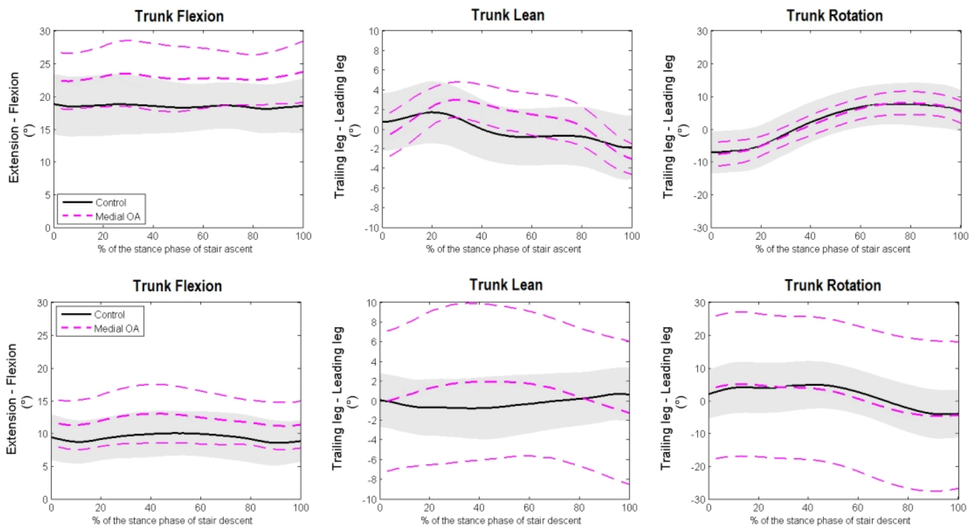

Figure 3 - Trunk kinematics relative to the ground reference frame in the sagittal (left), frontal (middle) and transversal (right) plane for step-over-step while ascending (above) and descending (below) stairs at controlled speed during stance phase, comparing healthy subjects and individuals with medial knee OA.

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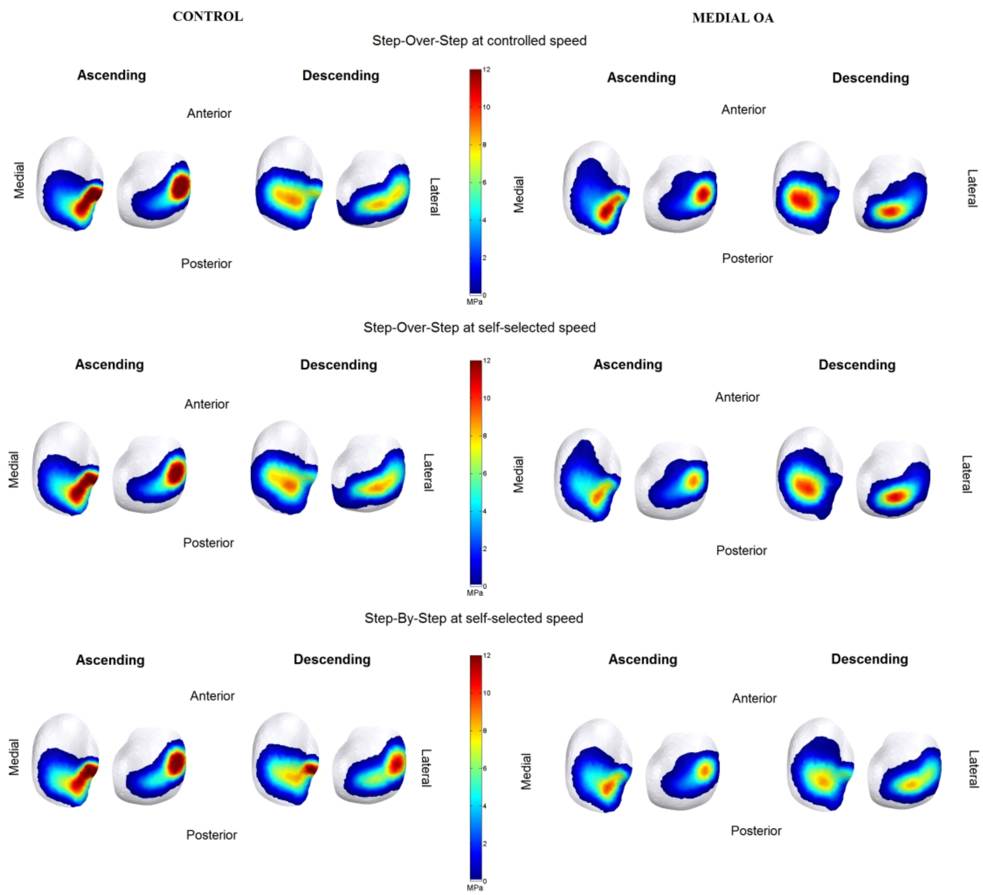

Figure 4 - Group-averaged contact pressure distributions (MPa) on the articular surfaces of medial tibial plateau at the time instant of the first peak medial KCF. To obtain these averaged contact pressure distribution maps, the average contact pressure was calculated for every triangle of the medial tibial surface mesh and presented on a representative surface model. Results are presented for step-over-step at controlled speed; step-over-step at self-selected speed and step-by-step, while ascending and descending stairs for the healthy group (on the left), and the medial knee OA group (on the right).

Table 1 - Characteristics of the groups: control and medial OA.

	Mean (SD)		p
	Control	Medial OA	(Control *vs* OA)
No. of subjects	8	5	-
No. of limbs	16	10	-
Age, *years*	51.0 (13.4)	52.8 (11.0)	.806
Body mass, *kg*	74.1 (13.7)	83.8 (14.8)	.255
Height, *m*	1.66 (0.10)	1.70 (0.11)	.489
KOOS score, %	96.7 (6.0)	42.3 (7.7)	**< .001**
KOOS pain score, %	96.5 (7.8)	41.1 (13.4)	**< .001**
KOOS function score, %	98.9 (2.0)	54.1(7.7)	**< .001**
R	0.968	0.876	
HOOS score, %	98.2 (4.6)	92.8 (10.4)	.214

Statistically significant differences ($p < .050$) between the two groups of subjects, evaluated by the independent *t*-test, are indicated in bold.

Function score indicates the level of function in activities of daily living (ADL).

R is the Person correlation coefficient between pain and function scores, in which 1 indicates a perfect correlation between the two parameters.

Table 2 – OA classification based on MRI and X-ray.

	Control		Medial OA	
	Lateral knee joint	Medial knee joint	Lateral knee joint	Medial knee joint
Cartilage score	0	0	0.6	1.8
BML	0	0	0.3	2
Osteophytes	0	0	1.2	1.6
K&L score	0		2-3 (4 out of 5)	

Table 3 – Stair walking speed during step-over-step at controlled speed (SOS CS) and at self-selected speed (SOS SS), and step-by-step (SBS) in patients with medial knee OA compared to controls.

			Mean (SD) Control	Mean (SD) Medial OA	p (Control *vs* OA)		p (Control)	p (OA)
Speed, *m/s*	Ascending	SOS CS	0.59 (0.02)	0.57 (0.04)	0.107	CS *vs* SS	.006	.031
		SOS SS	0.53 (0.08)	0.49 (0.12)	0.364			
		SBS	0.36 (0.04)	0.38 (0.03)	0.203	SOS *vs* SBS	< .001	.009
	Descending	SOS CS	0.60 (0.03)	0.56 (0.08)	0.154	CS *vs* SS	.180	.107
		SOS SS	0.57 (0.09)	0.49 (0.11)	0.057			
		SBS	0.34 (0.05)	0.36 (0.04)	0.303	SOS vs SBS	< .001	≤ .001

Statistically significant differences (p < .050) between the two groups of subjects, evaluated by the independent t-test, are indicated in bold.

Statistically significant differences (p < .050) between strategies (CS *vs* SS, and SOS *vs* SBS) within each group of subjects, evaluated by the paired-sample t-test, are indicated in bold.

Table 4 – Maximum contact pressures (*MPa*) at the peak medial KCF (MKCF) comparing the two groups of subjects and *p*- values comparing activities into the groups.

<table>
<tr><th colspan="2" rowspan="2"></th><th colspan="2">Mean (SD)</th><th rowspan="2">p
(C0 vs OA)</th></tr>
<tr><th>Control
(16 legs)</th><th>Medial OA
(10 legs)</th></tr>
<tr><td rowspan="2">SOS CS</td><td>Ascending</td><td>24.1 (12.1)</td><td>16.0 (6.1)</td><td>.092</td></tr>
<tr><td>Descending</td><td>15.8 (5.6)</td><td>14.2 (4.6)</td><td>.598</td></tr>
<tr><td rowspan="2">SOS SS</td><td>Ascending</td><td>24.4 (11.7)</td><td>13.9 (4.6)</td><td><u>.004</u></td></tr>
<tr><td>Descending</td><td>15.7 (7.1)</td><td>13.8 (4.6)</td><td>.317</td></tr>
<tr><td rowspan="2">SBS</td><td>Ascending</td><td>24.4 (12.6)</td><td>14.7 (4.6)</td><td><u>.035</u></td></tr>
<tr><td>Descending</td><td>16.1 (5.9)</td><td>11.4 (3.3)</td><td><u>.013</u></td></tr>
<tr><td rowspan="2">Ascending</td><td>p
(SOS SS vs SOS CS)</td><td>.717</td><td>.093</td><td rowspan="4"></td></tr>
<tr><td>p
(SOS SS vs SBS)</td><td>.877</td><td>.059</td></tr>
<tr><td rowspan="2">Descending</td><td>p
(SOS SS vs SOS CS)</td><td>.959</td><td>.445</td></tr>
<tr><td>p
(SOS SS vs SBS)</td><td>.877</td><td><u>.007</u></td></tr>
</table>

Statistically significant differences ($p < .050$) in maximum contact pressures between the two groups of subjects, evaluated by Mann-Whitney-U test, are indicated in bold.
Statistically significant differences ($p < .050$) in maximum contact pressures between strategies within each group of subjects, evaluated by Wilcoxon matched-pair test, are indicated in bold.
SOS CS, SOS SS and SBS correspond to step-over-step at controlled and self-selected speed, and step-by-step, respectively.

PART I -Knee Model

The multibody knee model was developed from MRI images of the right knee from a 23 years old female subject (height = 1.65 m, mass = 61 kg) with no history of chronic knee pain, injury, or surgery.

Anatomical reference frame orientations were established for each bone using an automatic algorithm based on geometric and inertial properties of the 3D segments [1, 2].

The tibiofemoral and patellofemoral joints were both modeled as 6 degree of freedom with deformable contact. The passive restraints of the knee joint are provided by the major knee ligaments and joint capsule, represented by 14 bundles of non-linear springs: superficial and deep medial collateral ligament (sMCL, dMCL), lateral collateral ligament (LCL), anteriomedial and posteriolateral anterior cruciate ligament (aACL, pACL), anteriolateral and posteriomedial posterior cruciate ligament (aPCL, pPCL), patellar tendon (PT), medial and lateral patellofemoral ligaments (MPFL, LPFL), popliteofibular ligament (PFL), posteriomedial capsule (pmCAP), the posterior capsule (CAP), and the iliotibial band (ITB). Each ligament bundle was represented by a discrete number of strands. Each strand was assumed as a non-linear stiffening spring at low strains ($\varepsilon < 0.06$), and having a linear stiffness at higher strains [3]. The ligament elastic modulus was assumed to be 125 MPa [4].

Tibiofemoral and patellofemoral cartilage geometry were segmented from MRI images (Mimics Innovation Suite, Materialise, Belgium). Tibiofemoral and patellofemoral cartilage contact pressures (*p*) acting between articulating surfaces were computed using an elastic foundation model [5], in which pressure is assumed to be a function of the depth of penetration between contacting cartilage surfaces.

$$p = -\frac{(1-v)E}{(1+v)(1-2v)}\ln\left[1-\frac{d}{h}\right], \quad (1)$$

with two additional equations resulting from the equilibrium of pressures in pairs of contacting triangles, and the equivalence of the sum of the individual surface penetration depths to the total penetration depth:

$$p_1 = p_2 \quad (2)$$

$$d_1 + d_2 = d \quad (3)$$

where E is elastic modulus, v is Poisson's ratio, d is the penetration depth and h is the combined thickness of the two cartilage surfaces. The system of equations (Eqs 2 and 3) is solved for each pair of contacting triangles (subscripts) in the cartilage meshes given the E, ν, and thickness of each cartilage geometry. Cartilage was assumed to have an elastic modulus of 5MPa [3], a Poisson's ratio of 0.45 [6] and represented by uniform cartilage thickness of 2mm over each surface (*i.e.* 4 mm total thickness).

The model included 44 musculotendon actuators spanning the right hip, knee and ankle [7].

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PART II - Comparison between control subjects and medial OA

S1 Figure - Total, medial and lateral knee contact forces (KCF) during step-over-step (SOS) at **controlled speed** while ascending (above) and descending stairs (below) comparing healthy subjects and individuals with medial knee OA. * indicates a significant difference between the groups.

S1 Table - Peak values of the total, medial and lateral KCF (×BW) during the stance phase of step-over-step at **controlled speed** (SOS CS), while ascending (ASC) and descending (DESC) stairs comparing between the control (C0) group and the medial OA (OA) group.

		Total (26)	Control (16 legs)	Medial OA (10 legs)	*p* (C0 vs OA)
ASC	P1	TKCF	4.49 (0.85)	3.17 (0.82)	.001
		MKCF	2.51 (0.28)	1.86 (0.54)	< .001
		LKCF	2.24 (0.81)	1.52 (0.36)	.015
	P2	TKCF	2.82 (0.65)	2.65 (0.53)	.492
		MKCF	1.56 (0.62)	1.52 (0.35)	.868
		LKCF	1.39 (0.43)	1.26 (0.44)	.454
DESC	P1	TKCF	3.26 (0.81)	2.72 (0.75)	.104
		MKCF	2.11 (0.57)	1.58 (0.41)	.019
		LKCF	1.28 (0.36)	1.34 (0.42)	.682
	P2	TKCF	4.33 (0.96)	3.43 (1.12)	.038
		MKCF	2.44 (0.54)	1.98 (0.65)	.063
		LKCF	2.11 (0.72)	1.58 (0.58)	.062

Statistically significant differences (p < .050) are indicated in bold and calculated by paired-samples t-test. KCF are expressed as mean (SD (BW), where SD is standard deviation. P1 and P2 correspond, respectively, to first and second peak of the stance phase.

S2 Figure - Knee flexion (left), adduction (middle) and rotation (right) moments during step-over-step (SOS) at **controlled speed** while ascending (above) and descending stairs (below) comparing healthy subjects and individuals with medial knee OA. * indicates a significant difference between the groups.

2 Table – Peak values of the KFM, KAM and KRM (BW*Ht) during stance phase of step-over-step at **controlled speed** (SOS CS) while ascending (ASC) and descending stairs (DESC).

		Total (26)	Control (16 legs)	Medial OA (10 legs)	p (C0 vs OA)
ASC	P1	KAM	0.017 (0.009)	0.016 (0.008)	.805
		KFM	0.070 (0.012)	0.050 (0.017)	**.002**
		KRM	-0.008 (0.006)	-0.006 (0.004)	.235
	P2	KRM	0.002 (0.003)	0.001 (0.003)	.633
DESC	P2	KAM	0.021 (0.008)	0.016 (0.007)	.119
		KFM	0.073 (0.015)	0.058 (0.018)	**.022**

Statistically significant differences (p < .050) are indicated in bold and calculated by paired-samples t-test. Knee moments are expressed as mean (SD (BW*Ht), where SD is standard deviation. P1 and P2 correspond, respectively, to first and second peak of the stance phase.

S3 Figure - Hip, knee and ankle kinematics at the sagittal (left), frontal (middle) and transversal (right) planes of rotation for step-over-step (SOS) while ascending stairs at **controlled speed** during stance phase comparing healthy subjects and individuals with medial knee OA.

S4 Figure - Hip, knee and ankle kinematics at the sagittal (left), frontal (middle) and transversal (right) planes of rotation for step-over-step (SOS) while descending stairs at **controlled speed** during stance phase comparing subjects and individuals with medial knee OA.

S3 Table - Trunk extension and bending angles (in degrees), at the time instant of the first peak MKCF during SOS at controlled speed.

		Control (16 legs)	Medial OA (10 legs)	*p*
SOS CS Ascending	Trunk Flexion Angles	18.43 (3.74)	24.45 (3.76)	<u>.001</u>
	Trunk Lean Angles	0.93 (2.82)	2.76 (1.38)	.069
SOS CS Descending	Trunk Flexion Angles	8.98 (3.43)	10.89 (3.11)	.166

	Trunk Lean Angles	0.09 (2.67)	0.73 (7.72)	.762

Statistically significances ($p < .050$) are indicated in bold and calculated by *t-test*.

Positive trunk flexion angles correspond to flexion of the trunk; positive trunk bending correspond to bending towards the leading leg.

SOS CS corresponds to step-over-step at controlled speed.

PART III - Comparison between strategies: SOS at controlled *versus* self-selected speed and SOS *versus* SBS

S5 Figure - Total, medial and lateral knee contact forces (KCF) in **healthy subjects** comparing step-over-step at controlled speed (SOS CS) and step-over-step at self-selected speed (SOS SS) while ascending (above) and descending (below) stairs. * indicates a significant difference between the two tasks.

S6 Figure - Total, medial and lateral knee contact forces (KCF) in **individuals with medial knee OA** comparing step-over-step at self-selected speed (SOS SS) and step-over-step at controlled speed (SOS CS) while ascending (above) and descending (below) stairs. * indicates a significant difference between the two tasks.

S4 Table - Peak values of the total, medial and lateral KCF (×BW) during the stance phase of step-over-step at controlled speed (SOS CS), step-over-step at self-selected speed (SOS SS) and step-by-step (SBS) while ascending and descending stairs for the control and medial OA groups comparing between tasks.

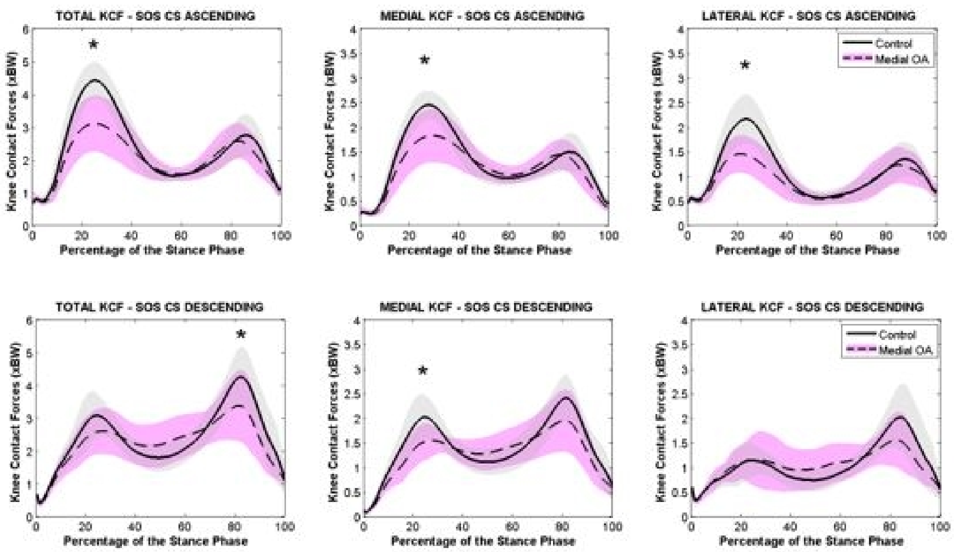
		ASCENDING					DESCENDING				
		SOS CS	SOS SS	SBS	*p* CS *vs* SS)	*p* (SS *vs* SBS)	SOS CS	SOS SS	SBS	*p* (CS *vs* SS)	*p* (SS *vs* SBS)
CONTROL	TKCF	4.49 (0.85)	4.41 (0.78)	4.56 (0.86)	.414	.182	4.33 (0.96)	4.20 (0.74)	4.44 (0.73)	.473	.087
	MKCF	2.51 (0.28)	2.61 (0.26)	2.64 (0.34)	.190	.672	2.44 (0.54)	2.44 (0.43)	2.43 (0.35)	.977	.797
	LKCF	2.24 (0.81)	2.04 (0.67)	2.17 (0.69)	**.009**	.066	2.11 (0.72)	1.92 (0.53)	2.31 (0.60)	.144	**.016**
MEDIAL OA	TKCF	3.17 (0.82)	2.78 (0.62)	2.94 (0.70)	**.007**	.101	3.43 (1.12)	3.29 (1.14)	3.48 (1.03)	.506	.215
	MKCF	1.86 (0.54)	1.64 (0.45)	1.81 (0.40)	**.024**	**.008**	1.98 (0.65)	1.90 (0.58)	1.95 (0.50)	.547	.657
	LKCF	1.52 (0.36)	1.32 (0.29)	1.36 (0.31)	**.002**	.425	1.58 (0.58)	1.52 (0.72)	1.73 (0.74)	.628	**.040**

Statistically significant differences ($p < .050$) are indicated in bold and evaluated by the paired-sample t-test. KCF are expressed as mean (SD (BW), where SD is standard deviation. KCF corresponding to the peak KCF of the different tasks, i.e., first and second peaks KCF for ascending and descending, respectively.

S5 Table - Trunk extension and bending angles (in degrees), at the time instant of the first peak medial KCF during SOS while ascending and descending stairs for the control and medial OA groups comparing between controlled and self-selected speed.

		ASCENDING			DESCENDING		
		SOS CS	**SOS SS**	***p* (CS *vs* SS)**	**SOS CS**	**SOS SS**	***p* (CS *vs* SS)**
CONTROL	**Trunk Flexion**	18.43 (3.74)	18.10 (3.26)	*.331*	8.98 (3.43)	0.00 (9.29)	*.753*
	Trunk Lean	0.93 (2.82)	0.83 (2.92)	***.004***	0.09 (2.67)	-1.00 (2.42)	*.254*
MEDIAL OA	**Trunk Flexion**	24.45 (3.76)	23.71 (3.31)	*.304*	10.89 (3.11)	11.50 (3.64)	*.157*
	Trunk Lean	2.76 (1.38)	3.06 (2.14)	*.602*	0.73 (7.72)	0.57 (2.01)	*.942*

Statistically significant differences ($p < .050$) in the trunk angles between strategies within each group of subjects, evaluated by the paired-sample *t*-test, are indicated in bold. Angles are expressed as mean (SD (º), where SD is standard deviation.



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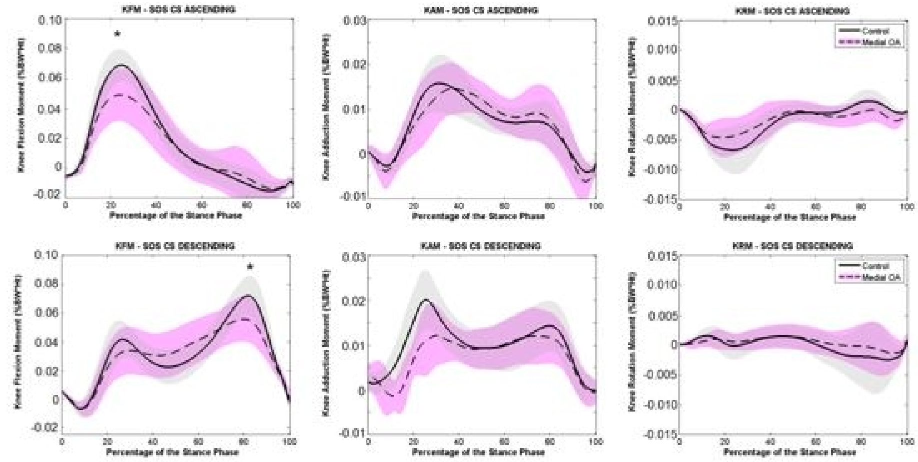

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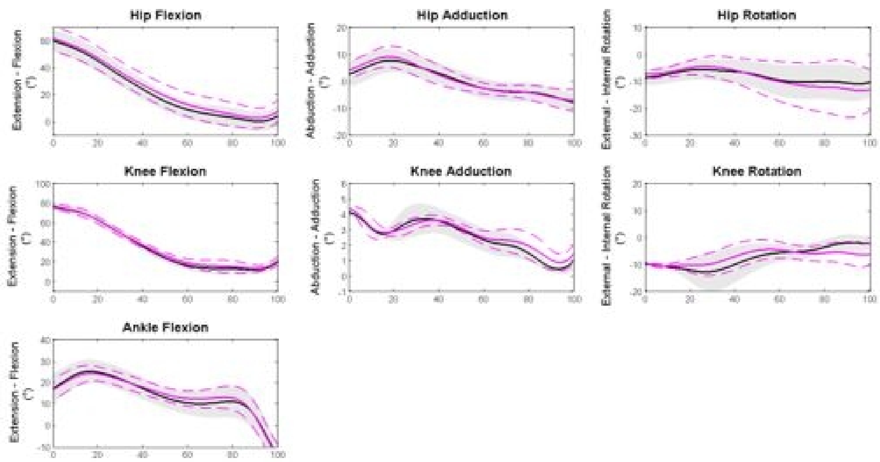

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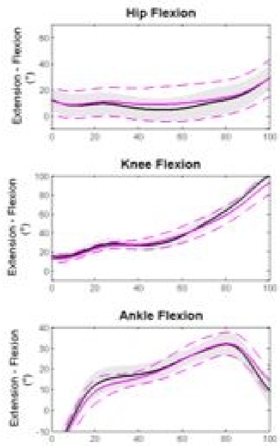

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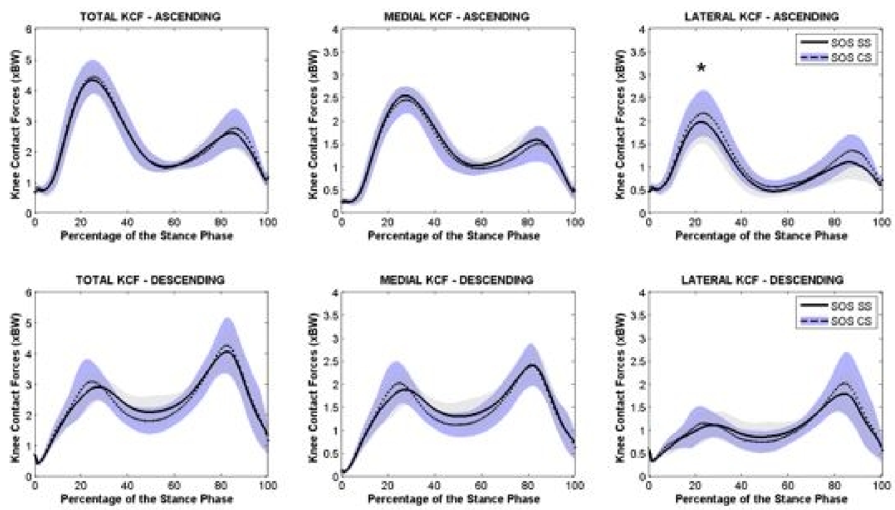

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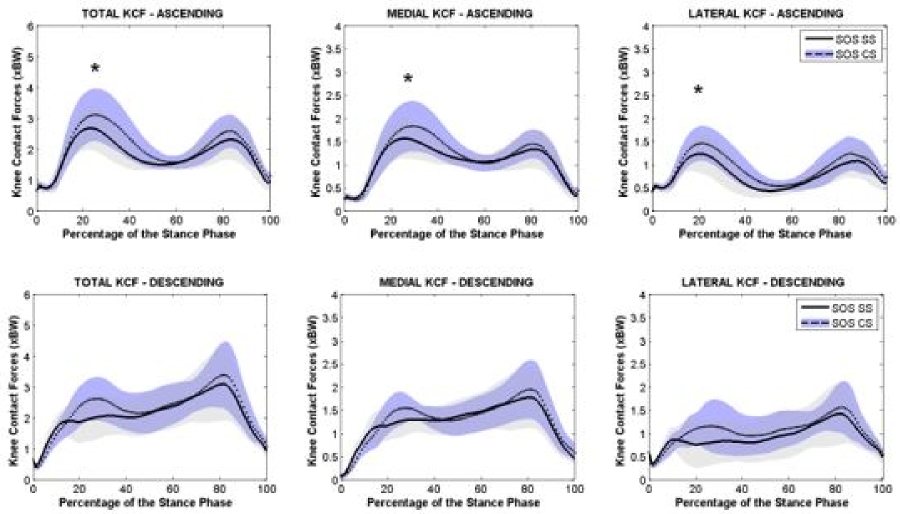

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