The effect of foot position during static calibration trials on knee kinematic and kinetics during walking.

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Abstract

Background

Gait analysis has been used extensively for computing knee kinematics and kinetics, in particular, in healthy and impaired individuals. One variable assessed is the external knee adduction moment (EKAM). Variations in EKAM values between investigations may be caused by changes in static standing position, especially foot placement angles which may increase or reduce any differences seen.

Purpose of the study

The current study aimed to explore the influence of static trial foot position on knee kinematic and kinetic variables during walking.

Methods

Twelve healthy male participants completed three different static standing trials; 1) 20-degrees toe-in, 2) 0 degree and 3) 20-degrees toe-out before walking at their own pace during a lower limb kinematics and kinetics assessment. First and second peak EKAM was compared between static foot position trials, as well other knee kinematic and kinetic outcomes. Repeated measures ANOVA was used with post hoc pairwise comparison to determine the differences between static foot position trials.

Results

The first peak of EKAM was significantly smaller in the 20^o toe-out angle, than the 20^o toe-in angle ($p = 0.04$ - 8.16% reduction). Furthermore, significant changes were found in peak knee kinematics and kinetics variables (adduction angle, external rotation angle, knee flexion moment external rotation moment, abduction angle and internal rotation angle) in the different positions.

Conclusion

Modification in static foot position between study visits may result in changes especially in the 1^{st} peak EKAM and other kinematics and kinetics variables during walking. Therefore, standardisation of static foot position should be utilised in longitudinal studies to ensure changes in EKAM are not masked or accentuated between assessments.

Keywords: Static standing position, 3D gait analysis, kinematics, kinetics, knee adduction moment

1. Introduction:

The gold standard for computing joint kinematics, joint kinetics, and spatiotemporal characteristics is gait analysis employing three-dimensional (3D) motion capture systems and force platforms. Such information may be valuable in complex clinical decision-making or when selecting the appropriate rehabilitation protocols [1]. Gait analysis has also been used to investigate musculoskeletal injuries and pathologies by comparing joint kinematics and kinetics with healthy individuals or tracking changes over time [2]. Furthermore, 3D gait analysis allows researchers to learn more about how individuals respond to treatments like bracing, conventional therapy, or surgery [3].

One of the pathologies that can impair gait is knee osteoarthritis (OA), which affects knee joint function and loading. Many investigations attempt to evaluate and measure knee function and joint load in individuals with knee OA [4]. The external knee adduction moment (EKAM) is a widespread surrogate loading measure, and recent investigations have found a relationship between EKAM and knee joint loading [5–7]. In addition, earlier investigations have shown that individuals with knee OA have higher EKAM than healthy people [5,8]. Moreover, a higher EKAM has been linked to the progression of knee OA [9,10]. In contrast, some studies have reported no difference in EKAM in individuals with knee OA when compared to healthy individuals [11,12]. This was supported by a systematic review and meta-analysis which found no consistent evidence regarding the EKAM between individuals with knee OA and those without [13]. Furthermore, identifying changes in EKAM between sessions is important, especially with biomechanical interventions to identify differences over time.

Many biomechanical factors have been investigated to explain the changes in EKAM in individuals with knee OA. It has been suggested that different walking approaches used by individuals with knee OA [14], progression of the disease [12,15], or methodological inaccuracies attributed to assessors such as marker misplacement, which could potentially explain the EKAM variation [16]. A previous study found that shifting the location of knee markers during a static trial caused almost a 25% variation in the sagittal plane knee moment [17] but no differences in the forntal (EKAM) or transverse plane. Moreover, toe-in and toe-out gait strategies have reduced EKAM while walking [18]. However, it is not yet clear how the variation of foot placement in static standing trials could impact the EKAM during the dynamic walking trials. Previous studies have investigated the effect of changing the static position (hip neutral, hip internal rotation, hip external rotation) on dynamic kinematics during walking using the conventional gait model [19]. Whilst this study only investigated hip rotation and knee frontal plane angle, they showed significant changes with changing the hip position during static trials.

Although the foot position in the static trial may affect the dynamic trial, previous research explored the EKAM during walking without providing a clear standardisation for the static trial. The few studies describing the standardised position during the static trial were limited to positioning the foot pointing forward with no internal or external rotation [20,21], or according to the individual assuming their comfortable position [21,22]. This is important in a repeated measures design and randomised clinical trials as alterations in standing foot position between visits may mask or accentuate the differences between groups and interventions.

Therefore, the current study aimed to explore the influence of static trial foot position on knee adduction moments and other knee kinematic and kinetic variables during walking. We hypothesised that changing the foot position while recording the static standing trial will affect the kinematic and kinetics of the knee data during walking.

2. Methods:

Twelve healthy male participants ranging in age from 18 to 33 years (mean age (SD): 23.5 ± 2.91 years, height: 1.73 ± 0.05 m, mass 70.93 ± 15.46 kg, body mass index (BMI): 23.55 ± 4.41 kg/m^2) participated in this study. The sample size calculation was based on the external knee adduction moment and via using a similar method to a previous study [23]. The g-power program was used with an effect size 0.52, alpha error 0.05 and 95 power.

Individuals were included if they could walk unassisted without any type of help and had a clear account of lower limb deformity or injury to their back, pelvis, or lower extremity in the previous three years. Individuals with neurological or neuromuscular disorders that may impair their ability to walk were not permitted to take part. Participants were invited by posters distributed around the University campus. The Hail University institutional review board granted ethical approval (H-2020-229), and each participant signed a consent form for participation and publication of results. The study was carried out at Hail University's gait lab. A motion analysis system of ten cameras (Vicon-Bonita infra-red motion cameras, Oxford Metrics, Oxford, UK) sampled at 100 Hz and two force plates (Advanced Mechanical Technology Incorporation (AMTI) force plate, Type OR67, Watertown, USA) sampling at 1000 Hz embedded flush in the ground were utilised in measuring lower limb kinematics and kinetics.

This study involved a within-subject repeated design where each participant was tested in three conditions. The three static positions used to analyse the same dynamic trials were with the foot positioned as follows: 1) 20-degree toe-in, 2) 0 degree, 3) 20-degree toe-out. (See Figure 1). Participants were instructed to bring their own everyday shoes.

Figure 1: Feet placement over the force plate during the three static positions (Left pictue = 0 degrees, Middle picture = Toe–in 20 degrees, Right picture = Toe–out 20 degrees

Following previous research [23-25], a custom protractor was created over the embedded force plate to guide the participant on their foot placement for the selected angles. The participant was instructed to align their heel and the second toe over the proposed angles. The intended foot placement angles in the current investigation were chosen based on previous knee loading studies [24,25]. An earlier investigation found that individuals with knee OA could walk with different toe-out angles of up to 36 degrees [24]. Furthermore, another study displayed that healthy individuals were able to adopt the increase and decrease in toe angle to each individual baseline by 13 to 25 degrees [25]. The proposed foot placement angles were randomised using block randomisation (randomization.com).

The Calibrated Anatomical System Technique (CAST) approach was used where passive retro-reflective markers with a diameter of 14.5 mm [26] were affixed to the lower limbs of the participants. To recognize the anatomical reference frame and centers of joint rotation, twenty-four markers positioned over the anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), iliac crests, greater trochanters, lateral femoral condyle, medial femoral condyle, lateral malleolus, medial malleolus, calcaneus, and 1st, 2nd, and 5th metatarsal heads. Additional four nonorthogonal markers affixed on rigid cluster plate were positioned on the anterior lateral side of each thigh and shank.

Participants then performed three static standing trials with 10 seconds hold in each trial and 30 seconds between trials according to the randomized block. During the static trial the participant was instructed to move their feet to the required positions which were clearly marked onto the floor, which were checked to ensure consistency between experimental conditions.. Afterwards, participants were instructed to walk at their own pace in order to gather a group of successful five walking trials. A successful dynamic trial was considered when the subject walked in a natural manner, landing the whole foot on the force plate at their predetermined own self-selected pace. Participants in all dynamic walking trials were unaware of the force plates to minimise force plate targeting.

The data obtained from the five successful walking trials and three static standing trials were processed using the Vicon Nexus version (Vicon Motion Systems, Oxford, UK), in which markers were identified and exported as a C3D file. Visual 3d (C-Motion, Inc., USA) program was utilized to create the biomechanical model using each static trial with the same five walking trials. A six-degree freedom model was utilised where all segments were modellled as rigid bodies. Anatomical frames were defined by markers (knee joint centers were considered as the mid-point between medial and lateral knee condyles markers while the ankle joint centers were considered as the midpoint between the medial and lateral malleoli markers) where right handed segment coordinate systems were defined. The hip joint centers were calculated using the regression model based on ASIS and PSIS markers [27]. Joint kinematics were calculated using an X–Y–Z Euler rotation sequence equivalent to the joint coordinate system [23].

An inverse dynamic approach was used to calculate the external joint moment via the Newton-Euler equation. Segment geometric properties and inertial were estimated for each participant [28,29]. The proximal segment was used to resolve the moments into the coordinate system as per previous research [30].

The data were interpolated and filtered with a low pass filter of 25 Hz for kinetics [31] and 6 Hz for kinematic [32]. Body mass was used to normalise the kinetic data. The primary outcome of the study was to account for the variation in EKAM (first peak, trough, second peak) between the same dynamic walking trials using different static standing positions. The first peak was calculated as the maxima between 0 and 33 percent of stance, the trough as the minima between 34 and 67 percent of stance, and the second peak as the maxima between 68 and 100 percent of stance [33,34].

Other kinetics and kinematics data were investigated as the knee sagittal plane angles (Initial contact, maximum early stance, maximum, minimum, ROM), knee frontal plane angles (maximum (adduction), minimum (abduction), ROM), knee transverse plane angles (maximum (Internal rotation), minimum (External rotation), ROM), Knee sagittal plane moments (flexion, extension), knee transverse plane moment (internal rotation, external rotation).

Microsoft Excel (version 15.29.1 for Mac) was used to create spreadsheets for the subsequent statistical analysis. Shapiro-Wilk test was used to check the data normality. The influence of using different foot position static trials on knee kinematic and kinetic variables during the same walking trials was examined using repeated measures ANOVA ($p < 0.05$) with pairwise comparison (Bonferroni adjustment) using Statistical Package for Social Sciences (SPSS) version 23 (SPSS Inc., USA). Partial eta squared was calculated when conducting one-way ANOVA and used to represent the effect size (0.01, 0.06, and 0.14 indicating small, medium, and large effect) [35].

3. Results:

The first peak of EKAM during walking was altered with the static standing foot position; where 20° toe-out was significantly lower than 20° toe-in ($p < 0.000$) with a 8.2 % reduction (Table 1, Figure 2). Other kinematic and kinetics variables which showed significant changes ($p<0.05$) were Peak knee adduction and abduction angles, Peak knee flexion moment, Peak knee internal and external rotation angles, and Peak knee external rotation moment (Table 2). Pairwise comparison (Table 3) showed that there were significant changes between all conditions in the knee adduction and abduction angles and knee flexion moments. The knee adduction angle reduced with an increased toe-out angle whilst the knee abduction angle and knee flexion moment increased with toe-out. A significant change was observed between 20° toe-in angle and other conditions in the knee external rotation and internal rotation angles with increase in the internal rotation angle as the static move toward toe-out and decrease in the external rotation angle as the static moved toward toe-out. In the external rotation moment, the differences were observed between 20° toe-out angle and 20° toe-in angle and between 20° toe-out angle and 0 degree with increase in the external rotation moment as the static trial moves toward external rotation.

Table 1: Mean (SD) for primary outcome external knee adduction moment (EKAM) for each foot static position angles during walking

Variables		Toe–in 20°	0°	Toe–out 20°	P-value	Effect size (Partial eta squred)
EKAM (Nm/kg)	1st peak	0.44 (0.11)	0.43 (0.1)	0.41 (0.1)	**0.01***	0.41
	Trough peak	0.26 (0.08)	0.26 (0.08)	0.26 (0.08)	0.65	0.03
	2nd peak	0.31 (0.11)	0.32 (0.11)	0.32 (0.1)	0.7	0.02

Bold=significant, *|significant *Toe–in 20° to Toe–out 20°.*

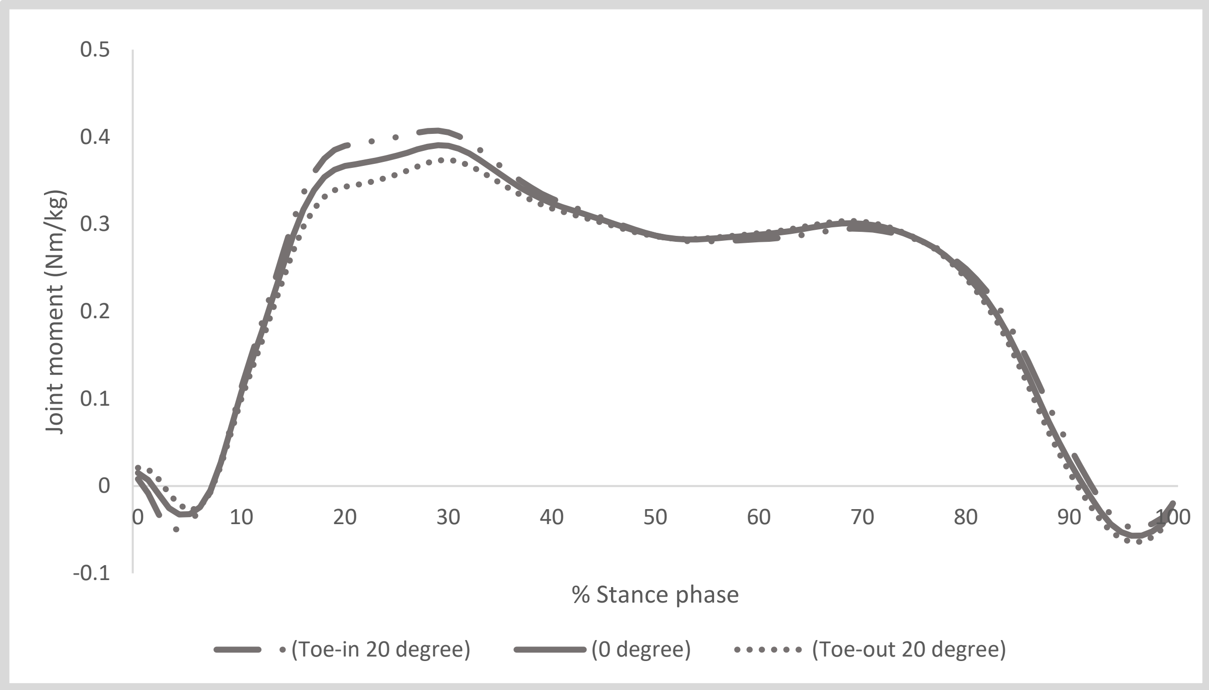

Figure 2: The figure displays the ensemble average of external knee adduction moment (EKAM, Postive indicate adduction) during stance phase walking with reference to the three static standing positions.

Table 2: Mean (SD) for other knee kinematics, and kinetics for each reference foot static standing angles during walking.

Variable	Value	Toe–in 20°	0°	Toe–out 20°	P-value	Effect size
Knee frontal plane angle (degree)	Maximum (Adduction)	9.39 (5.35)	6.27 (4.57)	3.95 (4.18)	**<0.01**#	0.66
	Minimum (Abduction)	-4.17 (4.6)	-7.01 (5.03)	-9.54 (4.84)	**<0.01**#	0.73
	ROM	13.56 (3.71)	13.29 (3.23)	13.49 (2.44)	0.86	0.006
Knee sagittal plane angle (degree)	Initial contact	3.51 (4.41)	3.43 (4.34)	2.96 (4.13)	0.12	0.424
	Maximum at early stance	10.45 (5.69)	10.5 (5.66)	10.07 (5.44)	0.25	0.12
	Maximum	67.61 (4.22)	67.5 (4.52)	67.07 (4.31)	0.07	0.21
	ROM	67.56 (3.58)	67.47 (3.55)	67.45 (3.47)	0.58	0.03
Knee sagittal plane moment (Nm/kg)	Extension moment	0.35 (0.1)	0.35 (0.1)	0.35 (0.1)	0.84	0.01
	Flexion moment	-0.33 (0.18)	-0.35 (0.18)	-0.37 (0.18)	**<0.01**#	0.63
Knee transverse plane angle (degree)	Maximum (Internal rotation)	-0.04 (5.89)	4.06 (6.33)	6.65 (3)	**<0.01*$**	0.57
	Minimum (External rotation)	-22.26 (6.32)	-17 (6.09)	-14 (5.65)	**<0.01*$**	0.65
	ROM	22.22 (5.27)	21.06 (5.25)	20.65 (5.54)	0.52	0.3
Knee transverse plane moment (Nm/kg)	Maximum (external rotation)	0.117 (0.03)	0.118 (0.03)	0.122 (0.03)	**0.01 $@**	0.33
	Minimum (internal rotation)	-0.076 (0.05)	-0.078 (0.05)	-0.077(0.05)	0.16	0.15

Bold=significant, *=significant between *Toe–in 20° to Toe–out 20°, $=significant between Toe–in 20° and 0°,@= significant between Toe–out 20 to 0°, #=significant between all conditions,*

Table 3: Pairwise comparison between condition for all kinematics and kinetics variables

Variable	Conditions	Conditions	Mean Difference	P-value	95% Confidence Interval for Difference	
					Lower Bound	Upper Bound
Knee frontal plane first peak adduction moment (EKAM)	Toe–in 20°	0°	0.02	0.06	-0.001	0.04
		Toe–out 20°	0.04	**0.04***	0.001	0.07
	0°	Toe–out 20°	0.02	0.11	-0.004	0.04
Knee flexion moment	Toe–in 20°	0°	0.02	**0.04***	0.00	0.04
		Toe–out 20°	0.04	**<0.01***	0.02	0.06
	0°	Toe–out 20°	0.02	**<0.01***	0.01	0.03
Knee frontal plane maximum angle (adduction)	Toe–in 20°	0°	3.12	**<0.01***	1.41	4.83
		Toe–out 20°	5.44	**<0.01***	2.41	8.46
	0°	Toe–out 20°	2.32	**0.03***	0.26	4.38
Knee frontal plane minimum angle (Abduction)	Toe–in 20°	0°	2.85	**<0.01***	1.51	4.18
		Toe–out 20°	5.37	**<0.01***	3.04	7.70
	0°	Toe–out 20°	2.52	**0.02***	0.49	4.56
Knee transverse maximum angle (Internal rotation)	Toe–in 20°	0°	-4.11	**<0.01***	-5.63	-2.59
		Toe–out 20°	-6.7	**<0.01***	-10.68	-2.71
	0°	Toe–out 20°	-2.59	0.35	-6.88	1.71
Knee transverse plane minimum angle (External rotation)	Toe–in 20°	0°	-5.26	**<0.01***	-7.48	-3.05
		Toe–out 20°	-8.27	**<0.01***	-12.68	-3.86
	0°	Toe–out 20°	-3	0.20	-7.20	1.19
Knee transverse plane maximum moment (external rotation)	Toe–in 20°	0°	-0.001	1.00	-0.006	0.004
		Toe–out 20°	-0.004	**0.02***	-0.008	-0.0006
	0°	Toe–out 20°	-0.003	**0.03***	-0.01	-0.0003

Discussion:

The purpose of this study was to understand how different foot positions assumed in static trials could affect the EKAM and other knee kinematic and kinetic characteristics during walking. Previously, investigations have examined the impact of assuming a variety of gait adaptations on biomechanical parameters in both healthy and individuals with knee OA [25,36,37]. Up to this date, this is the only study investigating the impact of performing different static foot placement positions on the EKAM and the other kinematics and kinetics outcomes. The results showed that when adopting a foot position from 20° toe-in to 20° toe-out static positions, the 1st peak EKAM value reduced, whilst the knee flexion moment and knee external rotation moment increased. This has implications for repeated assessments in individuals where the differences either may be accentuated or even reduced which potentially could be a reason for the lack of consensus in EKAM findings in a recent systematic review [13].

The different toe-in and toe-out foot positions during the static standing trials had a moderate effect on the magnitude of the first peak of EKAM. In terms of the EKAM, only the first peak significantly decreased with the toe-out foot position compared to 20° toe-in static foot position with an 8.2 % reduction. This reduction in EKAM (8.2%) is similar to previous intervention studies where the first peak of EKAM reduced by 5.2% to 9.1% with the use of LWI [30,38]. Moreover, a study that compared individuals with knee OA (severe) to those who were healthy showed an increase in 1st peak EKAM by 11.4% which is close to the current study value [12].

Therefore, these differences could have been attributed to different static foot positions rather than a treatment effect or an impairment effect.

The reduction in 1st peak EKAM was observed alongside increases in knee flexion and external rotation moments. It has previously been reported that medial knee loading is not solely attributed to EKAM, but moments in the other planes [20]. Our findings suggest different static foot positions could lead to altered sagittal and transverse moments which could influence findings from intervention and comparison when assessing knee joint loading.

The differences in EKAM may be attributed to the differences in knee adduction angles between the different static foot position angles. Despite these differences in knee frontal plane and transverse plane maximum and minimum angles no changes in ROM were observed between static foot positions suggesting a shift in graphs. Whereby a toe-in static foot position resulted in a shift towards greater knee adduction and external rotation angles. These changes in knee kinematics are likely to lead to considerable alterations of knee joint moment arm, i.e., the knees move close to the line of the GRF and potentially diminish the external GRF moment arms during walking [39]. A previous study showed that the mechanical axis was the best predictor for first peak EKAM which supports the current study findings [40]. Consequently, the static foot position is considered as one of the mechanical aspects that impact the first peak EKAM and should be taken into account when measuring the knee adduction moment during walking.

A previous systematic review highlighted the four main sources of error affecting the calculation of the joint moment in clinical investigations including 1- kinematic measurement and processing, 2- GRF measurement and processing, 3- joint model parameters determination and 4- inertial parameters estimation [41]. Any source of error in calculating 3D gait analysis variables (kinematic or kinetic) may lead to inappropriate clinical interpretation therefore inaccurate clinical decisions based on the amount of error. This highlights the importance of identifying any source of error or variation in order to be avoided to achieve accurate clinical interpretation to the data. Our findings highlight the need to standarise the static position when investigating the effect of a treatment on knee moments in the frontal, sagittal, and transverse planes.

Therefore, the impact of static foot position is likely to either further exaggerated or decrease the EKAM further when examining knee moments in healthy individuals or individuals with OA which could lead to misleading results during intervention studies. Therefore, longitudinal studies should control the foot position during the static trial in order to eliminate its effect on EKAM since an 8.2% reduction appears to be within the range of reductions seen with interventions.

Thewlis et al in a study determined the effect of using different knee marker locations (proximal and distal to the knee) to define the anatomical coordinate system during the static standing trial. Three marker configurations were utlised; medial and lateral condyles (FC), medial and lateral epicondyles (FE), and finally uring tibial ridges (TR). A significant change was found in the knee joint centre location in relation to the ankle joint centre in all planes. The study also showed significant changes in knee flexion and extension moments during walking. Although the study showed a difference in the 1st peak EKAM between the different marker configuration during static trial (FE= 0.35(0.16) Nm/kg , FC= 0.27 (0.07) Nm/kg, TR= 0.33 (0.08) Nm/kg) this was not significant.

Limitations and further directions

There are several limitations to the present study that should be acknowledged. One of these limitations is that the study was confined to young fit lean male participants, thus future studies need to be conducted on female and populations with knee OA populations to address the variability across the demographic spectrum. One more potential limitation to this study is that the EKAM was chosen as the primary outcome for the present study because it is a widely used measure of medial knee loading. There are various approaches to estimating knee load, such as employing an instrumented joint replacement to measure medial compartment compressive force or using quantitative musculoskeletal modeling. Despite this, research employing these approaches is limited due to their intrusive nature, high cost, complexity, and/or time-consuming nature. Importantly, strong relationships with internal medial knee joint contact force justify the use of EKAM as an excellent non-invasive approach to evaluate medial knee load. Finally, the present study was restricted to one dynamic activity (walking), therefore further studies are required to quantify EKAM across several activities such as running and stair negotiation to be more representative of everyday activities. The current study used the proximal segment to resolve the moment into the coordination system whereas different results may be seen when resolving to the distal segment. However, as this was a repeated measures analysis and the same subjects acted as their own control, it would be expected that the results would follow the same pattern.

Conclusion:

The findings of this study clearly demonstrate that variations in static standing foot position significantly affects the knee joint in the first peak of EKAM, flexion moment, knee external rotation moment, adduction angle, abduction angle, and external and internal rotation angles. The changes in the first peak of EKAM may be related to changes knee adduction angles and the subsequent shift of the knee joint moment arm. This implies that the static foot position is one of the mechanical factors that potentially affect the first peak EKAM and should be considered when conducting longitudinal studies to eliminate such effect. Future studies should standardise the foot position during the static trial to reduce such effects which may mislead the interpretation of kinematic and kinetic of the knee joint.

Conflict of interest: All authors have no conflict of interest

References:

[1] R. Smith, M. Chepisheva, T. Cronin, B.M. Seemungal, Diagnostic approaches techniques in concussion/mild traumatic brain injury: Where are we?, in: Neurosensory Disord. Mild Trauma. Brain Inj., Elsevier, 2019: pp. 247–277. https://doi.org/10.1016/B978-0-12-812344-7.00016-9.

[2] T. Balasukumaran, U. Gottlieb, S. Springer, Spatiotemporal gait characteristics and ankle kinematics of backward walking in people with chronic ankle instability, Sci. Rep. (2020). https://doi.org/10.1038/s41598-020-68385-5.

[3] W.S. Müller, B., Wolf, S. I., Brüggemann, G.-P., Deng, Z., McIntosh, A. S., Miller, F., & Selbie, Handbook of human motion, Springer Berlin, 2018.

[4] J. Favre, B.M. Jolles, Gait analysis of patients with knee osteoarthritis highlights a pathological mechanical pathway and provides a basis for therapeutic interventions, EFORT Open Rev. (2016). https://doi.org/10.1302/2058-5241.1.000051.

[5] L.E. Thorp, D.R. Sumner, J.A. Block, K.C. Moisio, S. Shott, M.A. Wimmer, Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis., Arthritis Rheum. 54 (2006) 3842–3849. https://doi.org/10.1002/art.22247.

[6] A. Trepczynski, I. Kutzner, G. Bergmann, W.R. Taylor, M.O. Heller, Modulation of the relationship between external knee adduction moments and medial joint contact forces across subjects and activities, Arthritis Rheumatol. 66 (2014) 1218–1227. https://doi.org/10.1002/art.38374.

[7] M. Wada, Y. Maezawa, H. Baba, S. Shimada, S. Sasaki, Y. Nose, Relationships among bone mineral densities, static alignment and dynamic load in patients with medial compartment knee osteoarthritis., Rheumatology. 40 (2001) 499–505.

[8] S.-C.C. Huang, I.-P.P. Wei, H.-L.L. Chien, T.-M.M. Wang, Y.-H.H. Liu, H.-L.L. Chen, T.-W.W. Lu, J.-G.G. Lin, Effects of severity of degeneration on gait patterns in patients with medial knee osteoarthritis, Med. Eng. Phys. 30 (2008) 997–1003. https://doi.org/10.1016/j.medengphy.2008.02.006.

[9] E.F. Chehab, J. Favre, J.C. Erhart-Hledik, T.P. Andriacchi, Baseline knee adduction and flexion moments during walking are both associated with 5 year cartilage changes in patients with medial knee osteoarthritis, Osteoarthr. Cartil. 22 (2014) 1833–1839. https://doi.org/10.1016/J.JOCA.2014.08.009.

[10] K.A. Marriott, T. Birmingham, R. Moyer, L. Kanko, R. Pinto, C. Primeau, R. Giffin, Association between high external knee adduction moment and increased pain during walking: within-limb comparisons in patients with medial compartment knee osteoarthritis, Osteoarthr. Cartil. 25 (2017) S112.

https://doi.org/10.1016/j.joca.2017.02.180.

[11] J.L. Astephen, K.J. Deluzio, G.E. Caldwell, M.J. Dunbar, C.L. Hubley-Kozey, Gait and neuromuscular pattern changes are associated with differences in knee osteoarthritis severity levels, J. Biomech. 41 (2008) 868–876. https://doi.org/10.1016/j.jbiomech.2007.10.016.

[12] A. Mündermann, C.O. Dyrby, T.P. Andriacchi, Secondary gait changes in patients with medial compartment knee osteoarthritis: Increased load at the ankle, knee, and hip during walking, Arthritis Rheum. 52 (2005) 2835–2844. https://doi.org/10.1002/art.21262.

[13] K. Mills, M.A. Hunt, R. Ferber, Biomechanical deviations during level walking associated with knee osteoarthritis: a systematic review and meta-analysis., Arthritis Care Res. (Hoboken). 65 (2013) 1643–1665. https://doi.org/10.1002/acr.22015.

[14] B.D. Street, W. Gage, The effects of an adopted narrow gait on the external adduction moment at the knee joint during level walking: evidence of asymmetry., Hum. Mov. Sci. 32 (2013) 301–313. https://doi.org/10.1016/j.humov.2012.08.007.

[15] K. Mills, M.A. Hunt, R. Ferber, No evidence of a consistent alteration in the external knee adduction moment during gait in individuals with knee osteoarthritis: a systematic review and meta-analysis, 2013. https://doi.org/10.1016/j.joca.2013.02.201.

[16] E. Szczerbik, M. Kalinowska, The influence of knee marker placement error on evaluation of gait kinematic parameters., Acta Bioeng. Biomech. 13 (2011) 43–46.

[17] D.J. Thewlis, D., Richards, J., & Bower, Discrepancies in Static Knee Marker Placement based on Different Anatomical Landmarks Produce Effects Greater than Previously Simulated., J. Appl. Biomech. 24 (2008) 185–190.

[18] K. Simic, M., Wrigley, T., Hinman, R., Hunt, M., & Bennell, Altering foot progression angle in people with medial knee osteoarthritis: The effects of varying toe-in and toe-out angles aremediated by pain and malalignment, Osteoarthr. Cartil. 21 (2013) 1272–1280. https://doi.org/10.1016/j.joca.2013.06.001.

[19] M.L. McMulkin, A.B. Gordon, The effect of static standing posture on dynamic walking kinematics: comparison of a thigh wand versus a patella marker., Gait Posture. 30 (2009) 375–378. https://doi.org/10.1016/j.gaitpost.2009.06.010.

[20] J.P. Walter, D.D. D'Lima, C.W. Colwell, B.J. Fregly, Decreased knee adduction moment does not guarantee decreased medial contact force during gait, J. Orthop. Res. 28 (2010) 1348–1354. https://doi.org/10.1002/jor.21142.

[21] M. Henriksen, T. Graven-Nielsen, J. Aaboe, T.P. Andriacchi, H. Bliddal, Gait changes in patients with knee osteoarthritis are replicated by experimental knee pain, Arthritis Care Res. 62 (2010) 501–509. https://doi.org/10.1002/acr.20033.

[22] N. Foroughi, R.M. Smith, A.K. Lange, M.K. Baker, M.A.F. Singh, B. Vanwanseele,

Dynamic alignment and its association with knee adduction moment in medial knee osteoarthritis, Knee. (2010). https://doi.org/10.1016/j.knee.2009.09.006.

[23] S.J. Khan, S.S. Khan, J. Usman, A.H. Mokhtar, N.A. Abu Osman, Combined effects of knee brace, laterally wedged insoles, and toe-out gait on knee adduction moment and fall risk in moderate medial knee osteoarthritis patients., Prosthet. Orthot. Int. 43 (2019) 148–157. https://doi.org/10.1177/0309364618796849.

[24] A. Chang, D. Hurwitz, D. Dunlop, J. Song, S. Cahue, K. Hayes, L. Sharma, The relationship between toe-out angle during gait and progression of medial tibiofemoral osteoarthritis., Ann. Rheum. Dis. 66 (2007) 1271–1275. https://doi.org/10.1136/ard.2006.062927.

[25] P.B. Shull, K.L. Lurie, M.R. Cutkosky, T.F. Besier, Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback, J. Biomech. 44 (2011) 1605–1609. https://doi.org/10.1016/j.jbiomech.2011.03.016.

[26] A. Cappozzo, F. Catani, U. Della Croce, A. Leardini, Position and orientation in space of bones during movement: anatomical frame definition and determination., Clin. Biomech. 10 (1995) 171–178.

[27] A.L. Bell, D.R. Pedersen, R.A. Brand, A comparison of the accuracy of several hip center location prediction methods., J. Biomech. 23 (1990) 617–21.

[28] D.W. Space, Requirements of the Seated Operator Geometrical, Kinematic, and Mechanical Aspects of the Bod, in: Dayton, OH: Wright-Patterson Air Force Base, 1995.

[29] E.P. Hanavan, A mathematical model of the human body, AMRL-TR-64-102.AMRL TR. (1964).

[30] R.K. Jones, M. Zhang, P. Laxton, A.H. Findlow, A. Liu, The biomechanical effects of a new design of lateral wedge insole on the knee and ankle during walking, Hum. Mov. Sci. 32 (2013) 596–604. https://doi.org/10.1016/j.humov.2012.12.012.

[31] E. Schneider, E.Y. Chao, Fourier analysis of ground reaction forces in normals and patients with knee joint disease., J. Biomech. 16 (1983) 591–601.

[32] D. Winter, Biomechanics and Motor Control of Human Movement, Fourth Edition, (2009). https://doi.org/10.1002/9780470549148.ch5.

[33] R.K. Jones, C.J. Nester, J.D. Richards, W.Y. Kim, D.S. Johnson, S. Jari, P. Laxton, S.F. Tyson, A comparison of the biomechanical effects of valgus knee braces and lateral wedged insoles in patients with knee osteoarthritis, Gait Posture. 37 (2013) 368–372. https://doi.org/10.1016/j.gaitpost.2012.08.002.

[34] O.W. Althomali, Influence of using knee sleeve and lateral wedge insole on knee loading among healthy individuals during stair negotiation., Gait Posture. 92 (2022) 103–109. https://doi.org/10.1016/j.gaitpost.2021.11.018.

[35] Andy field, Discover statistics using SPSS, fifth ed., SAGE Publications Ltd, London, 2018.

[36] S.K. Lynn, P.A. Costigan, Effect of foot rotation on knee kinetics and hamstring activation in older adults with and without signs of knee osteoarthritis., Clin. Biomech. 23 (2008) 779–786. https://doi.org/10.1016/j.clinbiomech.2008.01.012.

[37] P.B. Shull, R. Shultz, A. Silder, J.L. Dragoo, T.F. Besier, M.R. Cutkosky, S.L. Delp, Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis, J. Biomech. 46 (2013) 122–128. https://doi.org/10.1016/j.jbiomech.2012.10.019.

[38] R.K. Jones, G.J. Chapman, L. Forsythe, M.J. Parkes, D.T. Felson, The relationship between reductions in knee loading and immediate pain response whilst wearing lateral wedged insoles in knee osteoarthritis, J. Orthop. Res. 32 (2014) 1147–1154. https://doi.org/10.1002/jor.22666.

[39] T.R. Jenkyn, M. a. Hunt, I.C. Jones, J.R. Giffin, T.B. Birmingham, Toe-out gait in patients with knee osteoarthritis partially transforms external knee adduction moment into flexion moment during early stance phase of gait: A tri-planar kinetic mechanism, J. Biomech. 41 (2008) 276–283. https://doi.org/10.1016/j.jbiomech.2007.09.015.

[40] D.E. Hurwitz, a. B. Ryals, J.P. Case, J. a. Block, T.P. Andriacchi, The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain., J. Orthop. Res. 20 (2002) 101–107. https://doi.org/10.1016/S0736-0266(01)00081-X.

[41] V. Camomilla, A. Cereatti, A.G. Cutti, S. Fantozzi, R. Stagni, G. Vannozzi, Methodological factors affecting joint moments estimation in clinical gait analysis: a systematic review, Biomed. Eng. Online. 16 (2017) 106. https://doi.org/10.1186/s12938-017-0396-x.