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Dispersion of Knee Helical Axes during Walking after Maximal versus Resistant Strength Training in Healthy Subjects

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Abstract: Knee joint stability can be estimated during functional tasks through the analysis of the helical axes (HAs) dispersion. The study aimed at investigating (1) the test–retest reliability of knee HAs dispersion during walking and (2) the effects of maximal versus resistant strength training on knee HAs dispersion during walking. Thirty healthy subjects (age: 22.6 ± 2.1 years) randomized into a maximal training (MT) group and a resistance training (RT) group underwent a 2-week quadricep–hamstring strength training at 90% or 30% of the maximum voluntary contraction, respectively. Participants walked on a treadmill with clusters of retro-reflective markers placed on thighs and shanks to detect knee kinematics with an optoelectronic system. Knee HAs dispersion was assessed using mean distance (MD) and mean angle (MA) at 1 week before training start, before and after the first training session, and before and after the last training session. Moderate to excellent reliability was found for MD and MA on the sagittal plane (ICCs ≥ 0.70). No differences over time were found for MD and MA between MT and RT. HAs dispersion indexes resulted in reliable parameters for the quantification of knee stability on the sagittal plane during walking. Maximal and resistance strength training induced no knee HAs dispersion changes during walking.

Keywords: knee joint; strength training; helical axis; walking; joint stability

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1. Introduction

The knee joint consists of the patellar-femoral and tibial-femoral joints and is characterized by poor congruence among its articular surfaces, leading to a continuous center of rotation displacement during movements [1]. In this scenario, in addition to passive structures, such as ligaments and joint capsule, neuromuscular control plays a pivotal role in ensuring knee stability through a minimization of the joint rotation center displacement [2]. The helical axes (HAs) technique allows for an in vivo quantification of knee rotation center displacement through the computation of the instantaneous rotation axes during a movement and the analysis of their dispersion [3,4]. Mean distance (MD) and mean angle (MA) represent parameters adopted to quantify HAs dispersion. In particular, MD describes the HAs displacement providing a quantification of center of rotation movements, whereas MA describes the HAs orientation, revealing the ability to maintain a single plane of motion during a motor task [5–7]. Interestingly, knee HAs can be computed through the application of specific algorithms to kinematic data acquired using an optoelectronic system, allowing for the study of HAs dispersion not only during open-chain knee flexion–extension movements but also during the performance of functional activities, such as walking [8,9].

In the case of walking, the highest HAs dispersion has been reported in the stance phase of the gait cycle, when the horizontal component of the ground reaction force generates a considerable translational external moment at the level of the knee and high accuracy in neuromuscular control is required to ensure joint stability [9]. This phenomenon agrees with studies assigning a key role to neuromuscular control in determining HAs behavior, which represents a measure of joint stability [4,6,7,10].

Strength training has been reported to influence joint stability by means of mechanical changes and modifications in neuromuscular control [11,12]. In this context, different intensity protocols, including maximal resistant (80–90% of one repetition maximum) and submaximal resistant (30–40% of one repetition maximum) voluntary contractions are adopted, inducing different mechanical loads and metabolic stresses aimed at generating neuromuscular and structural adaptations [13–15]. When considering strength gains, both types of training have been reported to induce benefits, but a superiority has been demonstrated in favor of maximal strength training [16,17]. Interestingly, these strength modifications often occurred in the absence of differences in terms of muscle hypertrophy, especially in the short term, where neuromuscular adaptations play a predominant role in strength amelioration [15,17]. In addition, neuromuscular adaptations have been reported after a single session of maximal or resistant training [13]. However, the effects of maximal versus resistant strength training at the level of the knee joints have been mainly addressed in terms of strength modulation, neuromuscular activity changes, and molecular adaptations [12,13,18,19]. No studies have investigated the effects of such training modalities on knee stability during common daily tasks in which accuracy in neuromuscular control is required, such as walking.

Against this background, the analysis of knee HAs dispersion may be useful to identify changes in knee joint rotation center kinematics induced by maximal versus resistant strength training, providing information for the application of these training modalities in clinical practice. However, it is worth acknowledging that variations in terms of knee HAs parameters during walking, found by Temporiti and co-workers, are few and may also be partially attributed to a physiological variability in human gait [9]. In fact, a stride-to-stride joint kinematic variability has been described during walking in healthy subjects and the need to investigate knee HAs reliability during such a task represents a mandatory step to assign a clinical relevance to obtained results [20]. Therefore, this study aimed to (1) assess the test–retest reliability of HAs dispersion indexes during walking in healthy subjects and (2) investigate the effects of maximal versus resistant strength training on HAs dispersion during walking in healthy subjects.

2. Material and Methods

2.1. Participants

The study has a double-arm randomized design. Thirty young healthy volunteers were enrolled according to the following inclusion criteria: age between 20 and 30 years and right lower limb dominance assessed via the question “If you would shoot a ball on a target, which would be your preferred kicking leg?” [21]. Exclusion criteria were a history of lower limb or back impairments, knee pain, traumas, or lower limb surgery within the last year. Participants were volunteers, and they were enrolled from among students and employees of our institute through a recruitment email. After the enrollment, participants were randomized into a maximal training (MT) group and a resistance training (RT) group through a computerized random-number generator. Eligibility was assessed by an independent researcher blinded to the randomization list to ensure allocation concealment. The study was carried out at the Motion Analysis Lab of the Humanitas Clinical and Research Center of Milan, Italy, between June and February 2022. All participants signed an informed consent form, and the study was approved by our Internal Ethical Committee for Human Investigation (protocol number CLF21/06).

2.2. Intervention

Participants performed a 2-week (5 sessions per week) quadriceps–hamstrings maximal or resistant strength training with the dominant lower limb. Based on the key role of neuromuscular control in ensuring joint stability, the aforementioned training posology was adopted to mainly focus on the effects of early phase neuromuscular adaptations on HAs dispersion [14,17]. Participants were asked to sit upright on the chair of an isometric dynamometer (COR1, OT-Bioelettronica, Torino, Italy) with the hands resting on the armrests, hips at 90° flexion and the dominant knee at 50° flexion, secured with inextensible seatbelts. The dynamometer rotational axis was aligned with the lateral femoral condyle, and the lever arm was fixed about 3 cm above the lateral malleolus using an inextensible band with Velcro straps. The MT group performed isometric contractions of quadriceps and hamstrings at 90% of the MVC, whereas the RT group performed isometric contractions of quadriceps and hamstrings at 30% of the MVC [14]. Five repetitions were performed for each group of muscles up to failure, alternating in sequence the contractions of quadriceps and hamstrings. Verbal instructions were provided before each training session, and real-time visual feedback on target and developed forces was provided to participants during the contractions [14,17]. Before each training session, participants had to report a level of fatigue equal to 0 on the modified Borg scale [22]. An experienced physiotherapist supervised all the training sessions.

2.3. Knee Kinematics Assessment

Participants were assessed by a blinded operator for knee kinematics during walking at 1 week before the training start (T0), before (T1-pre) and 5 min after (T1-post) the first training session, and before (T2-pre) and 5 min after (T2-post) the last training session (Figure 1). During each evaluation session, participants were asked to walk barefoot on a treadmill for 120 s at a comfortable speed, previously estimated using the 10-m walking test and maintained unchanged for all the assessments [23]. Subjects were instructed to look straight in front of them and to not lean their hands on the treadmill during walking [24].

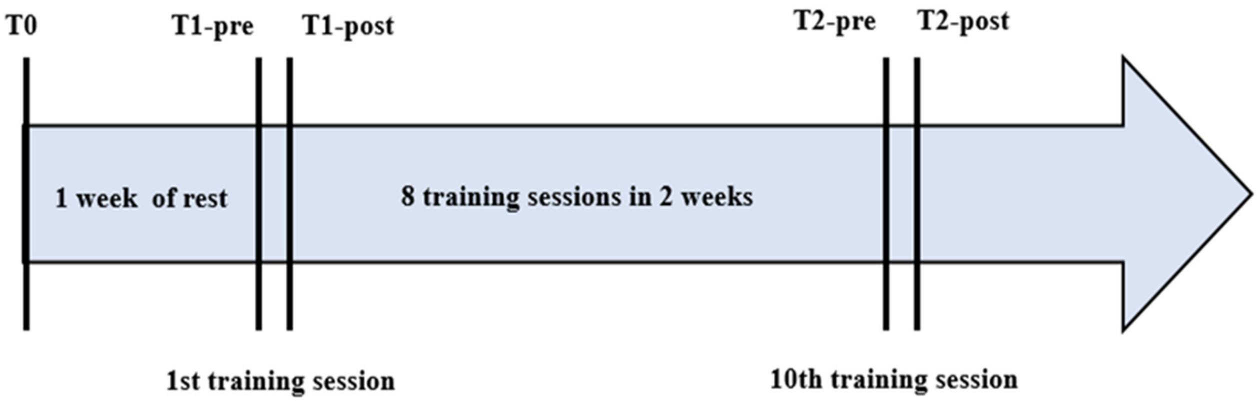

Figure 1. Representation of study design.

Knee kinematics during walking was detected using an optoelectronic system (BTS SMART-DX, BTS, Milano, Italy) equipped with 8 infrared cameras (sampling rate 100 Hz) placed around the treadmill in the standard position of a rectangular room. Two clusters of five retro-reflective markers were placed on the lateral surface of the thigh (equidistant between the greater trochanter and the lateral condyle of the femur) and shank (equidistant between the head of the fibula and the lateral malleolus) of the dominant and non-dominant limbs to identify femurs and tibias. The clusters were fixed through inextensible bands secured with Velcro straps to limit skin motion artifacts during walking. Two additional markers were placed bilaterally on the heel and on the 5th metatarsal head to identify the gait cycle phases [9].

Raw marker data were filtered using a fourth-order low-pass Butterworth filter (cut-off 4 Hz). Lower limb clusters were used to define thighs and shanks as rigid bodies, and the movement of the shank was normalized with respect to the thigh in order to compute their mutual positions at each timeframe as a composition of a rotation and translation around a fixed axis (HA) [25,26]. HAs were computed every 10° of knee motion along the sagittal plane, and their dispersion was described using MD and MA parameters, according to previous studies [3,6,26]. Gait cycle was defined as the timeframe between two consecutive heel-strike events, determined when the markers on the heels assumed the minimum value along the vertical axis of the global reference system [27]. In addition, the knee range of motion (RoM) was described through Euler angles and using the XYZ sequence, as recommended by the International Society of Biomechanics [28]. Furthermore, knee motion was divided into four phases based on kinematics along the sagittal plane: (1) flexion from 95% of the previous gait cycle to 10% of the subsequent gait cycle, (2) extension from 10% to 40% of gait cycle, (3) flexion from 40% to 70% of gait cycle, and (4) extension from 70% to 95% of gait cycle [9]. The first and the last five steps were removed to avoid artifacts, 80 consecutive steps were analyzed for each participant, and MD and MA were computed for each phase and described in reference to sagittal, frontal, and transverse planes adopting the methodology described by Temporiti and co-workers [9]. Finally, the percentage of HAs passing through the sagittal, frontal, and transverse planes during the four gait phases was also reported. Data were analyzed using MATLAB (MathWorks Inc., Natick, MA, USA) for Windows.

2.4. MVC Assessment

All participants were assessed for MVC at T1-pre and T2-pre after walking assessment and before the scheduled training session. The measurements were conducted in the same biomechanical condition of the training and using the same dynamometer (COR1, OT-Bioelettronica, Torino, Italy) by an experienced operator blinded to group allocation. Participants were asked to perform two MVCs interspaced by 3 min of rest. In particular, they were asked to exert force until reaching the maximum and hold that force level for 5 s. Standardized instructions and incitements were delivered to participants. Force signals were sampled at 100 Hz and, subsequently, filtered using a 0–33 Hz bandwidth. The highest force peak among the two trials was used for data analysis, and data were collected and processed using the software OT BioLab (OT-Bioelettronica, Torino, Italy) for Windows.

2.5. Statistical Analysis

All measurements were checked for normality using the Shapiro–Wilk test, and continuous variables were described as the mean and the standard deviation, while categorical variables were reported as proportions.

Test–retest reliability was investigated as agreement between T0 and T1-pre data (in terms of MD, MA, HAs percentage, and RoM) for the dominant and non-dominant limbs on each plane of motion during the four gait phases. Intraclass correlation coefficient (ICC, 2,k) was used to investigate relative reliability and interpreted as poor (lower than 0.50), moderate (between 0.50 and 0.75), good (between 0.75 and 0.90), and excellent (greater than 0.90) [29]. Absolute reliability was investigated using the standard error of measurement (SEM), calculated as SEM = SD $\sqrt{1\text{-ICC}}$, with SD representing the standard deviation of the mean difference between T0 and T1-pre trials.

When considering treatment effects, two-tailed *t*-tests for independent samples or chi square tests were used to assess between-groups differences in terms of participants' characteristics and outcome measures at baseline. Subsequently, mixed-model ANOVA (a 4 × 2 design for MD, MA, HAs percentage, and RoM and a 2 × 2 design for MVCs) with time as within-subject variable and group as between-subjects variable was used to investigate between-groups differences over time in terms of outcome measures. In the case of significant interactions or main effects, post hoc analysis with Bonferroni correction was performed to assess between-groups differences at each timepoint and within-group differences over time. The statistical level of significance was set at $\alpha = 0.05$, and the analysis was carried out using the software SPSS 28.0 for Windows (IBM Corp, Armonk, NY, USA).

3. Results

All participants (mean age 22.6 ± 2.1 years; 16 women and 14 men; mean height 173.2 ± 13.7 cm; mean weight 64.6 ± 10.4 kg) completed the training and the evaluation sessions correctly. No adverse events occurred during the training period, and the mean duration of each training session in RT and MT groups was 21.2 ± 2.8 min and 5.5 ± 1.2 min, respectively.

3.1. Test–Retest Reliability

MD resulted in good to excellent reliability on the sagittal plane (ICC from 0.75 to 0.90; SEM from 0.20 to 0.44 cm), poor to moderate reliability on the frontal plane (ICC from 0.23 to 0.63; SEM from 1.29 to 2.56 cm), and poor to good reliability on the transverse plane (ICC from 0.47 to 0.76; SEM from 0.62 to 1.34 cm).

MA revealed moderate to good reliability on the sagittal plane (ICC from 0.70 to 0.88; SEM from 0.76° to 2.79°), poor to good reliability on the frontal plane (ICC from 0.32 to 0.79; SEM from 3.72° to 7.75°), and moderate reliability on the transverse plane (ICC from 0.62 to 0.70; SEM from 3.40° to 4.89°).

HAs percentage showed good to excellent reliability on the sagittal plane (ICC from 0.78 to 0.90; SEM from 2.27% to 7.46%), poor to good reliability on the frontal plane (ICC from 0.42 to 0.86; SEM from 1.87% to 4.69%), and good to excellent reliability on the transverse plane (ICC from 0.79 to 0.84; SEM from 2.60% to 6.62%).

RoM revealed good to excellent reliability on the sagittal plane (ICC from 0.86 to 0.92; SEM from 0.63° to 1.15°), poor to moderate reliability on the frontal plane (ICC from 0.35 to 0.57; SEM from 0.92° to 3.79°), and moderate to excellent reliability on the transverse plane (ICC from 0.62 to 0.92; SEM from 0.50° to 2.68°) (Table 1).

Table 1. Test–retest reliability (ICC with 95% confidence interval and SEM) for mean distance (MD), mean angle (MA), percentage of HAs (%), and range of motion (RoM) on the sagittal, frontal, and transverse planes during the four gait phases.

Plane of Motion	MD (cm)				
	Phase	**T0**	**T1-pre**	**ICC (95% CI)**	**SEM**
Sagittal	95–10%	3.2 ± 0.9	3.3 ± 1.0	0.75 (0.56–0.84)	0.44
	10–40%	2.8 ± 1.4	2.9 ± 1.2	0.86 (0.77–0.92)	0.34
	40–75%	2.5 ± 0.8	2.7 ± 0.9	0.87 (0.79–0.92)	0.19
	75–95%	2.5 ± 1.0	2.6 ± 1.0	0.90 (0.83–0.94)	0.20
Frontal	95–10%	5.5 ± 1.8	5.5 ± 2.2	0.23 (−0.56–0.74)	2.56
	10–40%	4.8 ± 2.1	5.2 ± 1.9	0.47 (−0.18–0.76)	1.73
	40–75%	3.4 ± 1.9	4.0 ± 2.2	0.63 (0.31–0.80)	1.29
	75–95%	3.3 ± 1.5	3.9 ± 2.2	0.50 (−0.04–0.76)	1.52
Transverse	95–10%	3.0 ± 1.5	3.2 ± 1.4	0.76 (0.60–0.86)	0.62
	10–40%	3.4 ± 1.5	3.7 ± 1.6	0.47 (0.07–0.70)	1.34
	40–75%	2.9 ± 1.6	3.4 ± 1.8	0.59 (0.30–0.76)	1.16
	75–95%	2.5 ± 1.2	2.8 ± 1.3	0.61 (0.29–0.78)	0.82
Plane of Motion	**MA (°)**				
	Phase	**T0**	**T1-pre**	**ICC (95% CI)**	**SEM**
Sagittal	95–10%	14.2 ± 5.7	14.0 ± 4.7	0.70 (0.49–0.82)	2.79
	10–40%	11.0 ± 4.4	12.3 ± 5.5	0.78 (0.63–0.87)	1.92
	40–75%	10.1 ± 3.4	10.8 ± 3.5	0.88 (0.79–0.93)	0.76
	75–95%	10.3 ± 3.6	11.1 ± 4.7	0.74 (0.56–0.84)	1.94

Table 1. *Cont.*

Plane of Motion	MD (cm)				
	Phase	**T0**	**T1-pre**	**ICC (95% CI)**	**SEM**
Frontal	95–10%	22.6 ± 8.6	19.6 ± 8.8	0.44 (-0.43–0.79)	7.75
	10–40%	20.5 ± 10.3	20.0 ± 9.3	0.79 (0.54–0.91)	3.72
	40–75%	15.1 ± 4.8	16.1 ± 8.5	0.42 (−0.11–0.69)	6.40
	75–95%	12.8 ± 5.5	17.2 ± 9.1	0.32 (−0.29–0.66)	7.75
Transverse	95–10%	12.8 ± 6.9	14.4 ± 7.4	0.70 (0.49–0.82)	3.77
	10–40%	16.3 ± 9.2	13.2 ± 6.7	0.64 (0.36–0.79)	4.89
	40–75%	15.6 ± 6.1	16.5 ± 6.7	0.70 (0.47–0.83)	3.40
	75–95%	14.4 ± 6.7	15.8 ± 7.2	0.62 (0.32–0.79)	4.44
Plane of Motion	**HAs Percentage (%)**				
	Phase	**T0**	**T1-pre**	**ICC (95% CI)**	**SEM**
Sagittal	95–10%	76.2 ± 17.0	73.2 ± 19.9	0.78 (0.63–0.87)	7.46
	10–40%	79.6 ± 23.7	78.9 ± 24.6	0.87 (0.78–0.92)	6.07
	40–75%	83.5 ± 13.0	84.8 ± 11.9	0.90 (0.84–0.94)	2.27
	75–95%	84.5 ± 14.6	86.5 ± 11.5	0.86 (0.76–0.91)	3.52
Frontal	95–10%	3.0 ± 6.7	2.2 ± 3.9	0.48 (0.13–0.69)	4.63
	10–40%	2.8 ± 5.1	3.4 ± 8.4	0.86 (0.76–0.91)	1.87
	40–75%	3.7 ± 5.2	4.3 ± 4.7	0.70 (0.50–0.82)	2.60
	75–95%	3.8 ± 5.5	3.0 ± 4.6	0.42 (0.03–0.65)	4.69
Transverse	95–10%	20.9 ± 14.7	24.6 ± 18.2	0.79 (0.65–0.88)	6.14
	10–40%	17.6 ± 21.7	17.8 ± 22.1	0.84 (0.73–0.90)	6.62
	40–75%	12.7 ± 12.1	11.2 ± 10.3	0.84 (0.74–0.91)	3.28
	75–95%	11.8 ± 13.1	10.6 ± 10.6	0.89 (0.89–0.93)	2.60
Plane of Motion	**RoM (°)**				
	Phase	**T0**	**T1-pre**	**ICC (95% CI)**	**SEM**
Sagittal	95–10%	11.0 ± 5.6	11.3 ± 5.7	0.89 (0.82–0.94)	1.15
	10–40%	10.1 ± 4.3	9.6 ± 3.9	0.86 (0.77–0.92)	1.05
	40–75%	57.8 ± 4.0	57.2 ± 4.1	0.92 (0.85–0.95)	0.63
	75–95%	58.6 ± 4.7	58.8 ± 4.9	0.89 (0.81–0.93)	1.02
Frontal	95–10%	1.8 ± 1.3	1.7 ± 1.2	0.57 (0.28–0.75)	0.92
	10–40%	1.5 ± 1.1	1.3 ± 1.0	0.35 (−0.08–0.61)	1.06
	40–75%	5.5 ± 4.1	4.9 ± 4.2	0.50 (0.16–0.70)	3.39
	75–95%	4.8 ± 3.8	4.8 ± 3.9	0.37 (−0.06–0.63)	3.79
Transverse	95–10%	5.4 ± 3.2	5.9 ± 3.3	0.92 (0.86–0.95)	0.50
	10–40%	3.4 ± 2.4	3.1 ± 2.4	0.79 (0.64–0.87)	0.93
	40–75%	5.6 ± 4.1	5.2 ± 3.7	0.62 (0.37–0.77)	2.51
	75–95%	7.8 ± 4.6	7.8 ± 4.3	0.65 (0.41–0.79)	2.68

3.2. *RT versus MT*

No significant between-groups differences were found in terms of participants' characteristics and comfortable walking speed (RT 4.3 ± 0.4 km/h, MT 4.1 ± 0.2 km/h, and $p = 0.12$) (Table 2).

Based on test–retest reliability results, HAs dispersion on the sagittal plane was considered. No significant time by group interactions or group or time effects were detected for MD, MA, HAs percentage, and RoM in the untrained lower limb. When considering the trained limb, no significant time by group interactions or time effect were found, while a group effect was detected for MD during the fourth phase ($p = 0.045$), for MA in the first ($p < 0.001$) and fourth ($p = 0.035$) phases, and for HAs percentage in the third phase ($p = 0.011$). Post hoc analysis revealed lower MD at T1-post ($p = 0.047$) and T2-pre ($p = 0.033$) in favor of the MT group during the first phase of the gait cycle (Table 3).

Table 2. Participants' characteristics for RT and MT groups (independent sample t-tests and chi square tests). Data are shown as the mean ± the standard deviation.

	RT Group (*n* = 15)	MT Group (*n* = 15)	*p*-Value
	Variables		
Age (years)	22.5 ± 1.4	22.8 ± 2.7	0.739
Weight (kg)	66.3 ± 15.2	63.0 ± 12.4	0.524
Height (cm)	173.5 ± 12.1	173.0 ± 8.8	0.905
Gender (M/F)	7 M/8 F	7 M/8 F	1.000
Dominant limb (R/L)	15 R/0 L	15 R/0 L	1.000

Abbreviations: **M**: male, **F**: female, **R**: right, **L**: left.

Table 3. Between-groups (RT and MT) changes over time (T1-pre, T1-post, T2-pre, and T2-post) for mean distance (MD), mean angle (MA), HAs percentage (%), and range of motion (RoM) in the four gait phases on the sagittal plane for the trained lower limb (4 × 2 mixed-model analysis of variance with Bonferroni post hoc analysis). Data are shown as the mean ± the standard deviation, and significant results are shown in bold text.

Outcomes	Phase	T1-Pre		T1-Post		T2-Pre		T2-Post		*p*-Value	*p*-Value	*p*-Value
		RT	MT	RT	MT	RT	MT	RT	MT	Group Factor	Time Factor	Group × Time Interaction
MD	95–10%	3.6 ± 1.2	3.2 ± 1.0	3.8 ± 1.3	3.4 ± 1.2	3.5 ± 1.3	3.0 ± 0.9	3.5 ± 1.1	3.3 ± 0.9	0.073	0.635	0.991
(cm)	10–40%	3.2 ± 1.5	2.8 ± 0.9	3.0 ± 1.5	3.1 ± 1.3	2.8 ± 1.3	2.7 ± 1.0	3.0 ± 1.3	2.6 ± 0.9	0.398	0.769	0.762
	40–75%	2.5 ± 1.0	2.8 ± 0.8	2.6 ± 1.1	2.7 ± 0.8	2.4 ± 1.0	2.5 ± 0.8	2.6 ± 0.9	2.6 ± 0.7	0.458	0.85	0.942
	75–95%	2.3 ± 1.0	2.7 ± 0.7	2.3 ± 0.6	2.4 ± 0.7	2.2 ± 0.8	2.5 ± 0.8	2.3 ± 0.8	2.5 ± 0.8	0.045	0.845	0.947
MA	95–10%	13.0 ± 3.2	11.8 ± 3.5	14.4 ± 3.1 **	11.9 ± 3.4	14.0 ± 4.0 **	11.0 ± 3.3	15.0 ± 5.2	12.1 ± 3.8	<0.001	0.625	0.778
(°)	10–40%	11.8 ± 5.2	9.8 ± 4.1	11.7 ± 5.3	11.2 ± 5.3	10.9 ± 4.5	9.6 ± 2.7	13.0 ± 6.6	10.2 ± 4.4	0.064	0.689	0.829
	40–75%	11.3 ± 4.3	10.2 ± 3.6	11.2 ± 4.8	10.3 ± 3.6	10.9 ± 4.6	10.0 ± 3.3	9.4 ± 3.6	9.8 ± 3.6	0.378	0.643	0.898
HAs	75–95%	9.2 ± 2.4	11.0 ± 3.6	9.7 ± 2.3	11.1 ± 3.2	9.6 ± 3.4	10.6 ± 2.3	10.6 ± 3.7	11.0 ± 2.6	0.035	0.766	0.832
percentage	95–10%	75.4 ± 16.6	74.9 ± 14.3	75.5 ± 13.8	72.2 ± 16.3	74.2 ± 19.2	76.7 ± 18.0	74.1 ± 21.3	70.2 ± 18.3	0.686	0.881	0.887
(%)	10–40%	82.2 ± 26.2	83.0 ± 21.4	79.3 ± 24.4	70.2 ± 29.2	80.8 ± 24.6	84.1 ± 17.4	76.7 ± 25.1	76.7 ± 23.7	0.78	0.487	0.773
	40–75%	92.0 ± 9.4	88.6 ± 8.2	90.5 ± 9.1	84.8 ± 9.2	91.6 ± 8.5	90.0 ± 6.9	92.9 ± 7.2	87.6 ± 8.9	0.011	0.48	0.77
	75–95%	92.5 ± 8.3	86.0 ± 13.7	93.6 ± 6.7	90.7 ± 8.6	89.0 ± 12.6	88.7 ± 10.2	88.4 ± 15.7	91.7 ± 7.0	0.42	0.659	0.355
	95–10%	10.7 ± 6.3	12.2 ± 4.1	10.9 ± 5.9	11.0 ± 4.2	11.1 ± 7.4	12.6 ± 4.2	10.3 ± 6.6	10.8 ± 4.4	0.369	0.818	0.951
RoM	10–40%	9.9 ± 4.3	9.7 ± 3.0	9.7 ± 4.4	8.9 ± 3.5	9.5 ± 4.2	10.5 ± 3.0	9.3 ± 3.9	9.3 ± 3.3	0.984	0.827	0.83
(°)	40–75%	57.8 ± 3.2	56.2 ± 4.9	58.6 ± 4.4	56.7 ± 4.8	57.1 ± 3.4	56.6 ± 5.3	57.4 ± 4.0	57.2 ± 5.0	0.183	0.918	0.856
	75–95%	58.5 ± 3.2	58.7 ± 5.1	59.8 ± 3.3	58.8 ± 5.1	58.7 ± 4.8	58.6 ± 5.9	58.6 ± 4.4	58.6 ± 5.2	0.817	0.922	0.957

** Significant difference from MT at the same timepoint.

Finally, MVCs revealed no time by group interaction or group effect, whereas a time effect ($p < 0.001$) was found for quadriceps (RT: T1-pre 39 ± 13.1 kg, T2-post 43.8 ± 16.2 kg, and $p < 0.001$; MT: T1-pre 38.8 ± 10.7 kg, T2-post 45.5 ± 10.8 kg, and $p < 0.001$) and hamstrings (RT: T1-pre 14.6 ± 5.5 kg, T2-post 17.5 ± 6.3 kg, and $p < 0.001$; MT: T1-pre 14.5 ± 3.8 kg, T2-post 18 ± 5.9 kg, and $p < 0.001$).

4. Discussion

The study aimed to investigate the test–retest reliability of knee HAs dispersion indexes during walking in healthy subjects and assess the effects of maximal versus resistant strength training on knee HAs dispersion during walking in healthy subjects. In our study, knee HAs dispersion indexes revealed moderate to excellent test–retest reliability on the sagittal plane, whereas reliability decreased for HAs parameters on the transverse and frontal planes. Moreover, a 2-week maximal or resistant strength training of the quadriceps and hamstring muscles induced no knee HAs dispersion changes during walking in healthy subjects.

This was the first study addressed to investigate the reliability of knee HAs dispersion parameters during walking, showing higher reliability on the sagittal plane. The current findings may be related to larger knee excursion along the sagittal plane during walking, which resulted in a large number of computed HAs and high reliability in the estimation of the dispersion indexes [30,31]. In fact, in agreement with previous studies, HAs were computed every 10° of knee motion along the sagittal plane, in order to adopt an angle step able to provide a good compromise between movement analysis resolution and error in HAs estimation [9,26]. Consequently, this technique implied the assignment of a limited number of HAs to the frontal and transverse planes, where the range of motion is limited. In fact, it is worth noting that reliability consistently decreased according to the percentage of computed HAs, where higher HAs percentage on a plane seems to lead to higher accuracy in HAs parameters estimation (sagittal > transverse > frontal) [31]. Moreover, higher intra-subject variability in terms of knee kinematics has been described in healthy subjects on the frontal and transverse planes, when compared to the sagittal plane. This phenomenon may represent an additional influencing factor on the reliability of HAs dispersion parameters, especially when a limited number of steps are considered [30,32,33].

Based on reliability results, HAs assigned to the sagittal plane were exclusively considered to investigate the effects of maximal versus resistant strength training on knee HAs dispersion. No between-groups differences in terms of HAs dispersion were found during walking in healthy subjects, except for MA, which was lower in the MT group after the first training session and at training end in the first phase of the gait cycle. During this gait phase, knee is slightly flexed and a horizontal component of the ground reaction force results in considerable translational external moment [34]. This timeframe is particularly demanding in terms of neuromuscular control, which plays a key role in contrasting this moment and ensuring stability [34,35]. When considering our results, higher controlled knee kinematics seemed to occur in the MT group after a single training session and at the training end. However, when considering training effects, no changes over time were detected and it is worth noting that a certain degree of heterogeneity was still present before training between the two groups. In addition, the magnitude of observed changes is close to the SEM of 2.8° described for MA in the first phase of the gait cycle in healthy subjects, leading us to consider the relevance of the observed differences questionable.

In agreement with previous studies, both MT and RT groups had similar improvements in muscle strength, as demonstrated by MVC values at the training end [15]. However, no changes over time were detected in terms of knee HAs dispersion. In this scenario, a walking task at a comfortable pace in healthy subjects may not be demanding enough to highlight possible differences in terms of knee stability after a 2-week strength training.

Some limitations of the current study need to be underlined. Participants underwent strength training at a pre-established percentage of MVCs, which were collected at the beginning of the training. Therefore, strength gains were not monitored during the training period and the training intensity might be partially overestimated during training progression, especially during the last sessions. Moreover, quadricep and hamstring neuromuscular activity was not collected during walking, hindering the opportunity to detect possible training-dependent muscular adaptations in the absence of HAs dispersion differences. Finally, responsiveness of HAs dispersion parameters have never been investigated and low sensitivity or the presence of a ceiling effect in determining the lack of between-groups differences cannot be excluded.

In conclusion, HAs dispersion indexes resulted in reliable parameters for the quantification of knee stability on the sagittal plane during walking in healthy subjects. Moreover, maximal and resistance strength training of the quadricep and hamstring muscles induced no knee HAs dispersion changes during walking in healthy subjects. The current findings may be useful for the interpretation of the magnitude of HAs dispersion changes in future studies and provide reference values for the investigation of these parameters in subjects with knee joint disorders undergoing strength training.

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