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Influence of wedges on lower limbs' kinematics and net joint moments during healthy elderly gait using principal component analysis

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ABSTRACT

The elderly are susceptible to many disorders that alter the gait pattern and could lead to falls and reduction of mobility. One of the most applied therapeutical approaches to correct altered gait patterns is the insertion of insoles. Principal Component Analysis (PCA) is a powerful method used to reduce redundant information and it allows the comparison of the complete waveform. The purpose of this study was to verify the influence of wedges on lower limbs' net joint moment and range of motion (ROM) during the gait of healthy elderly participants using PCA. In addition, discrete values of lower limbs' peak net moment and ROM were also evaluated. 20 subjects walked with no wedges (control condition) and wearing six different wedges. The variables analyzed were the Principal Components from joint net moments and ROM in the sagittal plane in the ankle and knee and joint net moments in frontal plane in the knee. The discrete variables were peak joint net moments and ROM in sagittal plane in knee and ankle. The results showed the influence of the wedges to be clearer by analyzing through PCA methods than to use discrete parameters of gait curves, where the differences between conditions could be hidden.

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1. Introduction

The elderly population is rising in number due to a longer life expectation (Johnson, 2011). The elderly are more susceptible to many disorders (Granacher, Muehlbauer, & Gruber, 2012) that alter the gait pattern and, as a consequence, they are more vulnerable to fall (Kirkwood, de Souza Moreira, et al., 2011; Marcus et al., 2012), which affect their health and independence. In this perspective, the gait analysis of the elderly healthy population seems to be important in order to constitute reference values for understanding the abnormal gait pattern, as well as to assess the influence caused by different interventions on it.

Wedge insoles are often used to correct altered gait pattern. This therapeutic approach has been described as a powerful tool for the compensation of small gait deviations. According to Kerrigan et al. (2002), the use of insoles can influence decisively the quality of walking. Based on the different parameters that constitute an insole (e.g. height, material and density, and its relation with the shoes (e.g. position), it is possible to build insoles adequate to individuals with different diseases (Kerrigan et al., 2002). However, there is no consensus about the influence of different wedge insoles on the typical gait pattern (Van Gheluwe & Dananberg, 2004). Some authors found differences in gait waveforms (Chiu & Shiang, 2007; Erhart, Mündermann, Mündermann, & Andriacchi, 2008; Schmalz, Blumentritt, Drewitz, & Freslier, 2006) while others concluded that these devices did not influence the gait pattern of healthy subjects (Kakihana et al., 2005; MacLean, Davis, & Hamill, 2006).

Most of the recent studies evaluating the gait pattern of elderly subjects presented results of kinematics (Chui & Lusardi, 2010; Kirkwood, Resende, et al., 2011), plantar pressure distribution (Kirkwood, de Souza Moreira, et al., 2011) or, combine kinematic and kinetic parameters (Russell & Hamill, 2011; Trombini-Souza et al., 2011). These results are commonly presented as parameters extracted from discrete points in the kinematic and kinetic curves (called in this study as traditional approach), generating a huge amount of data, that sometimes is difficult to interpret (Chui & Lusardi, 2010). This approach relies on the definition of discrete parameters that are subjective, and it becomes difficult to extract the same values of all temporal waves, especially in the presence of pathologies (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007). A significant barrier to the clinical use of gait information is the complexity and large amount of data generated in biomechanical evaluations (Chau, 2001).

In the last two decades the interpretation of gait data was improved by different methods of multivariate analysis (Deluzio & Astephen, 2007; Jones, Holt, & Beynon, 2008; Muniz & Nadal, 2009; Olney, Griffin, & McBride, 1998; Sadeghi et al., 2002a). Principal component analysis (PCA) is a powerful method used to reduce redundant information and it allows the comparison of the complete waveform, explaining much of the variance in the data with relatively few factors, or Principal Components (PCs) (Sadeghi et al., 2002a). This approach may bring new insights about changes in gait pattern, to help clinicians to identify gait deviations and then to decide the best intervention for the patients. Therefore, the purpose of this study was to verify the influence of wedges on lower limbs' kinematics and net joint moment during walking of healthy elderly participants using PCA. In addition, the discrete gait parameters were also calculated in order to verify which approach (traditional or PCA) were more successful to determine changes in gait pattern. We hypothesized that using PCA the influence of wedges on lower limbs' kinematics and net joint moment waveforms will be observed. We also hypothesized that PCA approach will be more successful to determine changes in gait pattern compared to the traditional approach of discrete parameters.

2. Methods

This is a repeated measure study with a convenience sample. Ethical approval was granted by the institution in which the research was carried out. All participants freely signed an informed consent agreeing to participate.

2.1. Participants

Considering that the population with no gait dysfunction presents up to 10% of asymmetry for the force parameters between limbs during gait (Herzog, Nigg, Read, & Olsson, 1989), we assumed a difference of 15% in the knee moment in the frontal plane for the differences to have clinical relevance. In this way, with the mean and standard deviation of 0.4 ± 0.08 (Nm/kg) from a previous study (Kerrigan et al., 2002), which investigated a sample with similar characteristics of the present study, the sample size estimation suggested a minimum of 18 participants to achieve the statistical significant level of 0.05 with power of 0.85 for comparison among conditions.

Only subjects older than 50 years practicing physical activity regularly were included in this study. Those subjects who showed any kind of limitation or pain during walking were excluded. Thus, 20 physically active participants, 14 females and 6 males (mean age of 68.3 ± 9.4 years old; mean body mass of 66.0 ± 9.0 kg, mean body height 1.61 ± 5.9 m) were enrolled in this study.

2.2. Instruments

A Simi Motion System® (SIMI Reality Motion Systems, Unterschleissheim, Germany) with four cameras recording at 50 Hz, placed at the corners of the walkway was used to collect kinematic data. A piezoelectric force plate (Kistler® Instruments AG, Winterthur, Switzerland) recording at 1000 Hz was used to record the ground reaction forces, and a FootScan pressure plate (RsScan®, Olen, Belgium) recording at 300 Hz positioned over the force plate was used to record the center of pressure trajectory. Anti-slip mat was used to avoid any gap between the pressure plate and the floor. All systems were synchronized.

2.3. Experimental setup

Nine reflective markers were placed on the following landmarks: 2nd metatarsal head, malleolus lateralis, calcaneus, femur lateral epicondyle, femur great trochanter, Iliac spine anterior and posterior superior, 5th lumbar vertebra. All participants freely walked wearing their own shoes over an 8 m walkway, in which the force plate was embedded in the middle. The participant's shoes were of the same type (athletic shoes). After a short adaptation walking over the walkway, each subject performed 21 trials walking at their self-selected speed. At first, they performed three trials with no intervention, which was labeled as control condition (CON). Then, they performed, in random order, more three trials with six different wedges inside both shoes.

The wedges were placed at three different plantar foot regions and they were made of polyurethane cushion in six different shapes (Fig. 1):

(1) Lateral one (1L): This wedge was 1 cm high and it was placed under the 5° metatarsal bone.
(2) Lateral two (2L): This wedge was 2 cm high and it was placed at the same position than 1L.
(3) Medial one (1M): This wedge was 1.1 cm high and it was placed under the medial longitudinal foot arch.
(4) Medial two (2M): This wedge was 2.2 cm high and it was positioned as 1M.
(5) Posterior one (1P): This wedge was 0.9 cm high and it was placed under the calcaneous bone.

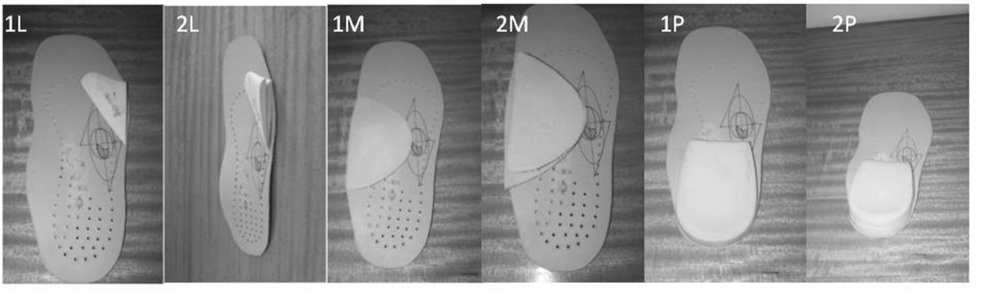

Fig. 1. Wedges conditions applied in the shoes. 1L: lateral one; 2L: lateral two; 1M: medial one; 2M: medial two; 1P: posterior one; 2P: posterior two.

(6) Posterior two (2P): This wedge was 1.8 cm high and it was placed at the same position than 1P.

Therefore, seven conditions were assessed: CON, 1L, 2L, 1M, 2M, 1P and 2P. We choose these wedges as they are the most used in our center of rehabilitation, and they are commonly (Chiu & Shiang, 2007; Erhart et al., 2008; Kakihana et al., 2005; MacLean et al., 2006; Schmalz et al., 2006).

2.4. Kinematics and kinetic data

The digitization and 3D reconstruction of images were performed in the *Dvideo v*.5.0 (Unicamp, Campinas, Brazil) videogrammetry system (Barros, Russomanno, Brenzikover, & Figueroa, 2006). The markers were tracked in a semi-automatic way and then, every marker in each frame was manually verified by one of the researchers. The kinetic data was processed in Simi Motion System® (SIMI Reality Motion Systems, Unterschleissheim, Germany) software and the plantar pressure data by Foot-Scan® 7 gait 2nd generation (RsScan®, Olen, Belgium) software. The integration of the data and muscle moments calculations was performed using Matlab 7.0 (MathWorks®, Massachusetts, USA).

We used the inverse dynamics model proposed by Vaughan, Davis, and O'Connor (1992) with the kinematic data processed after 3D calculation through the solidification procedure described by Cheze, Fregly, and Dimnet (1995). The kinematics (joint angles) of the ankle, knee and hip in the sagittal plane, and the joint net moment of the ankle in the sagittal plane, and knee in the sagittal and frontal planes along the central step of the trials were obtained. Gait speed was also calculated the first derivative of the iliac crest marker displacement.

To reduce the effect of random noise, the data was filtered using a Butterworth filter. The ground reaction force data with cutoff frequency of 4 Hz in 10th order and the kinematic data with cutoff frequency of 10 Hz in 4th order. The signals were interpolated and resampled in order to obtain 100 points (variables), providing one variable for each percent of stance phase. In the case of variables that also analyze the swing gait phase (kinematic data), 140 variables were obtained.

2.5. Traditional approach

To represent discrete values generally assessed in gait analysis we calculated the range of motion (the highest minus the lowest joint angle during the analyzed step) of ankle ($\text{Ankle}_{\text{ROM}}$), Knee ($\text{Knee}_{\text{ROM}}$) and Hip ($\text{Hip}_{\text{ROM}}$) in the sagittal plane; and the peak net moment (the highest net moment value during the analyzed step) in the ankle in the sagittal plane ($\text{Ankle}_{\text{SAG}}$), and knee in the sagittal (Knee_{SAG}) and frontal planes (Knee_{FRT}).

2.6. Principal component analysis

PCA was performed in joint net moment and kinematics waveforms according to Deluzio et al. (1997). In summary, the aim of PCA is to summarize the information contained in 100% of stance phase waveform in a smaller number of components that explain the greater variance through linear combinations from those variables, by considering each 1% in time axis as one variable (100 variables – matrix X and A below), and to represent the full waveform by a smaller number of components (PC model – matrix Z) that explain most of the variance through linear combinations from those variables (Jolliffe, 2004). The PC models defined by the equation $Z = U^tX$, where U are the eigenvectors of the covariance matrix of X (matrix S). U_n is calculated by the equation $SU_n = \lambda U_n$ where λ are the 100 eigenvalues. PCs are arranged in decreasing order in such a way that the first PC accounts for as much of the variability in the data, and each succeeding component accounts for much of the remaining variability as possible (Daffertshofer, Lamoth, Meijer, & Beek, 2004). In conclusion, each PC is a representation of a transformation of the data x, into z uncorrelated variables. Therefore, these uncorrelated variables are not redundant and explain different aspects of data variance. Finally, the practical implications of each PC may be drawn by analysis of the portion of the waveform in which it is relevant (description below).

2.6.1. PC model calculation

In this work, the generated PC model (Matrix Z) is a 100×100 matrix, determined by the product of U (100 eigenvectors) and X (100 columns representing the stance phase in CON condition). From matrix Z, the first three columns (PC1, PC2 and PC3) were retained for analysis, since according to a previous study (Jolliffe, 2004) the first 3 PCs contain the most variability explained. This procedure was performed 6 times, one for each of the analyzed outcomes (i.e. PC model from the (i) ankle joint net moment in the sagittal plane (ii) knee joint net moment in the sagittal plane, (iii) knee joint net moment in the frontal plane, (iv) ankle joint angles in the sagittal plane, (v) knee joint angles and (vi) hip joint angles in the sagittal plane).

2.6.2. PC scores calculation

The PC score values (s_n) are obtained by applying the equation $s_n = ZA$, where A is all the matrix (n subjects times 100 columns of data, equivalent to matrix X, but for experimental conditions) containing the data from the condition where the model is expected to be applied, (i.e. wedge conditions). This procedure generates a vector (1 line times n subjects) of data where each waveform (each subject in each wedge condition) is represented by a number (score).

In summary, for the determination of the influence of the wedges, the PC model (matrix Z) was developed based on the gait pattern of the subjects walking without any wedge (CON condition) and then this model was applied to the subjects walking with the wedge conditions (matrix A). The mean waveform of the three valid trials of each participant in each condition was used, and the PC score values (internal product from Matrix Z PC1, PC2 and PC3 to matrix A) for each subject in each condition were retained for analysis (PC1, PC2 and PC3 scores for the six waveforms, totalizing 18 PC score values per subject with each wedge). In the last phase, load vectors were calculated by normalizing the PC models (matrix Z) between -1 and 1 according to Jones et al. (2008). After normalization, a threshold of ±0.71 was adopted to consider a load vector from one variable as relevant, and then to attribute a meaning for this PC (Knapp & Comrey, 1973). It means that a variable only with values above these loadings, have a biomechanical interpretation in that portion of the curve. (Jones et al., 2008). For more information about PCA and its application previous studies are recommended (Jones et al., 2008; Sadeghi et al., 2002b).

2.7. Statistical procedures

The normality of the distribution of the scores (PC1, PC2 and PC3 from the Ankle_{SAG}, Knee_{SAG}, $\text{Knee}_{VAL,}$ Ankle, Knee and Hip kinematics), peak net moments (PeakAnkle_{SAG}, PeakKnee_{SAG}, PeakKnee_{FRT}) and ROM (Ankle_{ROM}, Knee_{ROM} and Hip_{ROM}) were tested by Kolmogorov–Smirnov test and the comparison between CON condition and wedges conditions (1L, 2L, 1M, 2M, 1P and 2P) was performed by the One Way ANOVA with a *post hoc* LSD. Also, the scores generated for each condition were ranked, and the 95% confidence interval (CI) was calculated in order to evaluate the range of the data inside this interval. The highest and lowest scores of these ranges were selected to highlight the main differences between conditions. These statistical procedures were made using SPSS (v.17; SPSS Inc., Chicago, IL) software with a significance level of $\alpha = 0.05$.

3. Results

All variables presented normal distribution. No differences were found among the mean gait speed in all tested conditions ($F = 2.078$; $p = .058$).

3.1. Traditional approach

Considering the discrete variables analyzed, there were no statistically significant differences between any conditions compared to CON for the peak net moment variables (Ankle_{SAG}, Knee_{SAG} and Knee_{FRT}), and for two range of motion variables (Knee_{ROM} and Hip_{ROM} – Table 1). Only the Ankle_{ROM} for 2L condition presented statistically significant difference from CON condition ($p < .001$).

3.2. Principal component analysis

In relation to the PCA approach, all wedge conditions were statistically significant different from CON in, at least, one of the six analyzed waveforms (Table 2). Using the PC model, five out of the six analyzed waveforms were statistically significant different when the wedge conditions were

Table 1
Discrete variables analyzed: (mean ± SD) for the lower limbs' moment peaks (PeakAnkle$_{SAG}$, PeakKnee$_{SAG}$, PeakKnee$_{FRT}$) and range of motion (Ankle$_{ROM}$, Knee$_{ROM}$ and Hip$_{ROM}$).

Variables	Wedges						
	CON	1L	2L	1M	2M	1P	2P
PEAK (Nm/BW)							
Ankle$_{SAG}$	2.5 ± 2.4	3.0 ± 2.4	6.0 ± 1.8	0.0 ± 1.8	7.0 ± 1.8	4.0 ± 1.8	2.5 ± 2.4
Knee$_{SAG}$	2.6 ± 2.3	2.8 ± 2.2	2.4 ± 2.4	2.7 ± 2.2	2.3 ± 1.4	0.0 ± 2.8	4.3 ± 4.4
Knee$_{FRT}$	7.7 ± 2.6	7.5 ± 2.3	6.7 ± 7.0	8.4 ± 4.0	7.7 ± 2.7	6.1 ± 2.8	7.9 ± 2.2
ROM (degrees)							
Ankle$_{ROM}$	40.8 ± 4.3	40.9 ± 4.3	53.0 ± 5.6*	35.6 ± 2.7	35 ± 1.7	41.1 ± 4.0	39.4 ± 4.7
Knee$_{ROM}$	50.4 ± 1.8	54.3 ± 7.0	55.8 ± 2.5	52.5 ± 2.3	5 4.5 ± 1.9	52.0 ± 2.7	50.0 ± 5.4
Hip$_{ROM}$	46.5 ± 1.5	49.7 ± 0.0	50.8 ± 2.0	46.7 ± 1.9	4 5.6 ± 1.3	48.6 ± 5.0	49.2 ± 2.7

CON: control condition; 1L: lateral one; 2L: lateral two; 1M: medial one; 2M: medial two; 1P: posterior one; 2P: posterior two.
* Statistically significant differences from CON group.

Table 2
PCA analysis: (mean ± SD) for PC1, PC2 and PC3 scores for the joint moments (Ankle, Knee sagittal, Knee frontal) and angle waveforms (Ankle, Knee and Hip).

Variables	Wedges						
	CON	1L	2L	1M	2M	1P	2P
Joint moments							
Ankle							
PC1	0.17 ± 0.92	0.13 ± 0.71*	0.56 ± 0.38*	0.16 ± 0.42	−0.19 ± 0.63*	0.01 ± 0.64	0.20 ± 0.94
PC2	0.24 ± 0.75	0.08 ± 1.26	−0.32 ± 0.90	0.05 ± 0.75	−0.04 ± 0.47	0.11 ± 0.82	−0.07 ± 0.94
PC3	0.09 ± 0.74	−0.14 ± 0.75	−0.12 ± 1.27	−0.12 ± 0.72	0.08 ± 0.87	0.01 ± 0.64	−0.18 ± 0.81
Knee sagittal							
PC1	0.17 ± 0.73	0.22 ± 0.77	−0.39 ± 0.92	0.00 ± 0.84	−0.09 ± 0.71	0.01 ± 0.88	0.46 ± 1.48
PC2	0.14 ± 0.85	−0.21 ± 0.96	0.25 ± 1.23	−0.36 ± 0.75	−0.15 ± 0.58	0.09 ± 0.98	0.19 ± 1.28
PC3	0.09 ± 0.62	0.20 ± 0.76	−0.01 ± 1.39	−0.16 ± 0.95	−0.28 ± 0.44	0.02 ± 0.52	0.85 ± 0.10
Knee frontal							
PC1	−0.26 ± 0.95	0.09 ± 0.90	0.29 ± 1.23*	−0.44 ± 0.76	−0.14 ± 0.57	0.17 ± 0.67*	−0.04 ± 0.61*
PC2	0.48 ± 1.01	−0.65 ± 0.91	−0.01 ± 1.04	0.14 ± 1.40	0.54 ± 0.88	0.34 ± 0.70	−0.34 ± 1.28
PC3	0.25 ± 0.9	0.39 ± 1.00	−0.07 ± 1.38	−0.74 ± 1.20	−0.62 ± 1.00	0.06 ± 0.93	0.53 ± 0.81
Joint kinematics							
Ankle							
PC1	0.31 ± 0.74	0.27 ± 0.68	−0.73 ± 0.93*	0.57 ± 0.24	0.52 ± 0.42	−0.09 ± 0.81	0.58 ± 1.26
PC2	−0.24 ± 0.85	−1.39 ± 0.66*	−0.77 ± 1.12*	−0.72 ± 0.99*	−0.01 ± 1.26	0.26 ± 1.19	−0.96 ± 0.91*
PC3	0.13 ± 0.99	−0.76 ± 0.68*	−0.96 ± 0.83*	−0.29 ± 0.35	−0.31 ± 0.66	−0.12 ± 1.18	0.20 ± 0.65
Knee							
PC1	−0.13 ± 0.89	1.02 ± 0.83*	0.83 ± 0.81*	0.58 ± 1.11	0.67 ± 1.02	0.48 ± 1.02	−1.05 ± 1.18
PC2	−0.13 ± 0.86	−0.05 ± 0.98	−1.41 ± 1.03	0.07 ± 1.46	−0.09 ± 1.35	−0.04 ± 1.23	−0.72 ± 1.22
PC3	0.01 ± 1.00	−0.01 ± 0.95	−0.49 ± 1.36	0.30 ± 0.16	0.11 ± 0.81	−0.26 ± 0.73	−0.19 ± 0.96
Hip							
PC1	0.00 ± 0.79	−0.30 ± 1.17	−0.06 ± 1.59	−0.94 ± 2.38	−0.45 ± 0.81	−0.36 ± 0.72	−0.17 ± 1.39
PC2	0.67 ± 1.36	−0.74 ± 2.36	−1.17 ± 1.15*	−2.24 ± 1.40*	−0.82 ± 1.35*	−0.91 ± 1.10*	0.07 ± 2.34
PC3	0.00 ± 0.98	−0.34 ± 1.09	−0.40 ± 1.37	−0.32 ± 1.57	−0.16 ± 0.88	−0.04 ± 1.10	0.33 ± 1.07

CON: control condition; 1L: lateral one; 2L: lateral two; 1M: medial one; 2M: medial two; 1P: posterior one; 2P: posterior two; PC: principal component.
* Statistically significant differences from CON group.

compared to CON: (i) ankle joint net moment in the sagittal plane, (ii) knee joint net moment in the frontal plane, (iii) ankle joint angles, (iv) knee joint angles and (v) Hip joint angles (Table 2). Only the knee joint net moment in the sagittal plane waveform seems not to be affected by the wedges and then, this waveform was not explored:

(1) Ankle joint net moment in the sagittal plane: The PC1 waveform was relevant and positive from 20% to around 80% of the stance phase (Fig. 2a). The conditions statistically different from CON in PC1 (Table 2) were 1L ($p = .01$), 2L ($p < .001$) and 2M ($p = .04$). The differences of 1L, 2L and 2M conditions compared to CON were in the same direction (lower values than CON). We can observe in the ankle moment in the sagittal plane PC1 waveforms with the highest and lowest scores lower values of joint net moment in 2L condition compared to CON where the PC1 was relevant (Fig. 2b). In PC2 and PC3 waveforms there were no statistically differences between the wedge conditions and CON.

(2) Knee joint net moment in the frontal plane: The conditions that presented statistically significant differences from CON were 2L ($p = .01$), 1P ($p < .001$) and 2P ($p = .02$). These differences were observed in PC1 scores (Fig. 3a), and in the 1P condition the highest differences were

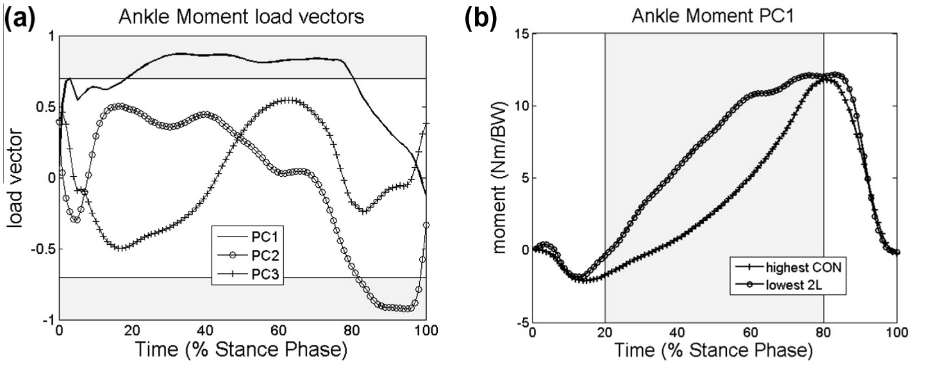

Fig. 2. Ankle moments in sagittal plane: (a) load vectors for PC1, PC2 and PC3; (b) highest and lowest scores in 95% confidence interval (CI). PC: principal component; CON: control condition; 2L: lateral two condition. Negative values represent ankle dorsiflexor moment, and positive values represent ankle plantarflexor moment. The grey area highlights the 0.71 threshold.

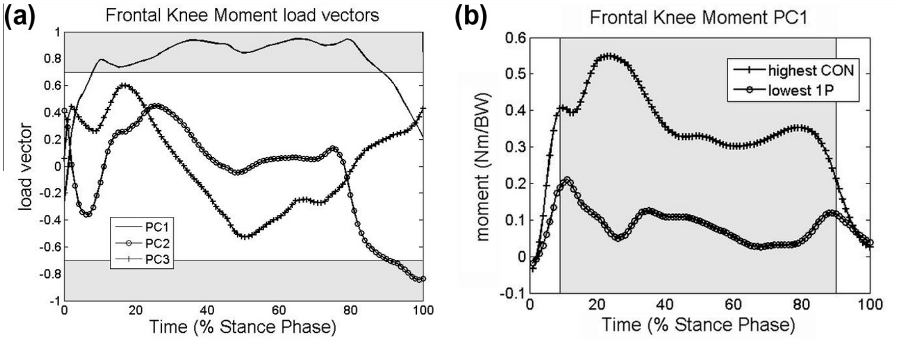

Fig. 3. Knee frontal moment: (a) load vectors for PC1, PC2 and PC3; (b) highest and lowest scores in 95% confidence interval. PC: principal component; CON: control condition; 1P: posterior one condition. Positive values represent knee valgus moment, and negative values represent knee varus moment. The grey area highlights the 0.71 threshold.

observed. The PC1 in all single support phase (from 10% to 90%) was relevant. The wedges reduced the knee joint net moment in the frontal plane (valgus moment) compared to walking with no wedges—CON (Fig. 3b).

(3) Ankle joint angles: The PC1, PC2 and PC3 were relevant (Fig. 4a) and, when the wedge conditions were compared to CON, their scores were statistically significant different (Table 2). The PC1 was relevant at the end of stance, beginning and end of the swing phase; the PC2 between 75% and 85% of the stance phase; whereas the PC3 showed its load vector relevant between 20% and 60% of the stance phase (Fig. 4a). In PC1, the 2L condition showed an increased range of motion at the end of gait cycle ($p < .01$—Fig. 4b), while the other wedges were similar compared to CON. In PC2, four conditions (1L, 2L, 1M and 2P) statistically influenced the ankle kinematics, among them, the 1L showed the highest differences compared to CON. This wedge (1L) increased the plantar flexion at the beginning of the toe off ($p < .01$—Fig. 4c). Finally, we observed in PC3 an increased dorsiflexion at the middle of the stance phase promoted by 2L ($p < .01$—Fig. 4d).

(4) Knee joint angles: Only PC1 identified statistically significant differences between the conditions and CON (Table 2). This PC was relevant at the beginning of the swing phase (Fig. 5a), in which the 1L and 2L conditions decreased the knee flexion (Fig. 5b).

(5) Hip joint angles: Hip kinematics was influenced by the wedges 2L ($p = .03$), 1M ($p < .01$), 2M ($p < .01$) and 1P ($p = .03$) in PC1 (Table 2). The PC1 (Fig. 6) was relevant during the whole stance phase and 20% of the swing phase (Fig. 6a). The wedges tended to decrease hip flexion and to increase hip extension (Fig. 6b).

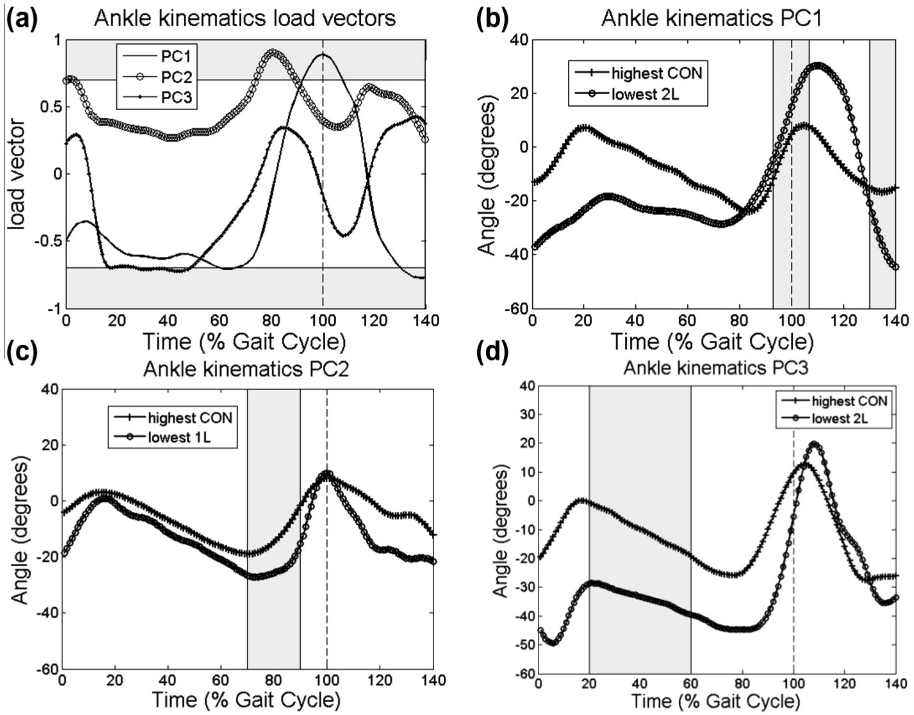

Fig. 4. Ankle range of motion in sagittal plane, in total gait cycle (stance and swing phase): (a) load vectors for PC1, PC2 and PC3; (b) highest and lowest scores in 95% confidence interval (CI) in Ankle$_{ROM}$ PC1; (c) highest and lowest scores in 95% CI in Ankle$_{ROM}$ PC2; (d) highest and lowest scores in 95% CI in Ankle$_{ROM}$ PC3. PC: principal component; CON: control condition; 1L: lateral one condition; 2L: lateral two condition. Positive values represent ankle plantar flexion, and negative values represent ankle dorsiflexion. The grey area highlights the 0.71 threshold.

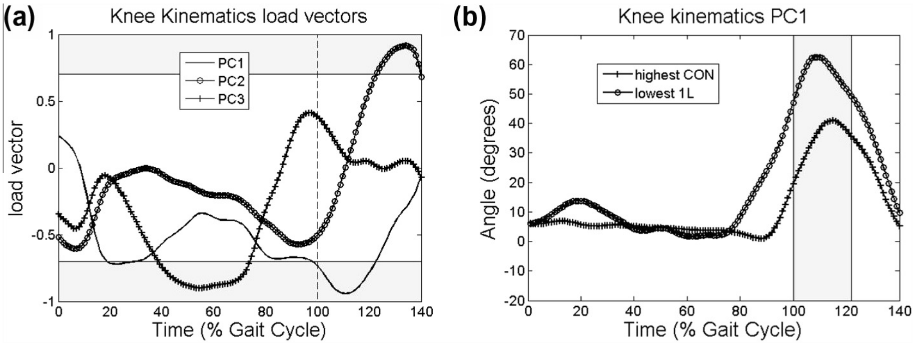

Fig. 5. Knee range of motion in sagittal plane: (a) load vectors for PC1, PC2 and PC3; (b) highest and lowest scores in 95% confidence interval. PC: principal component; CON: control condition; 1L: lateral one condition. Positive values represent knee flexion, and negative values represent knee extension. The grey area highlights the 0.71 threshold.

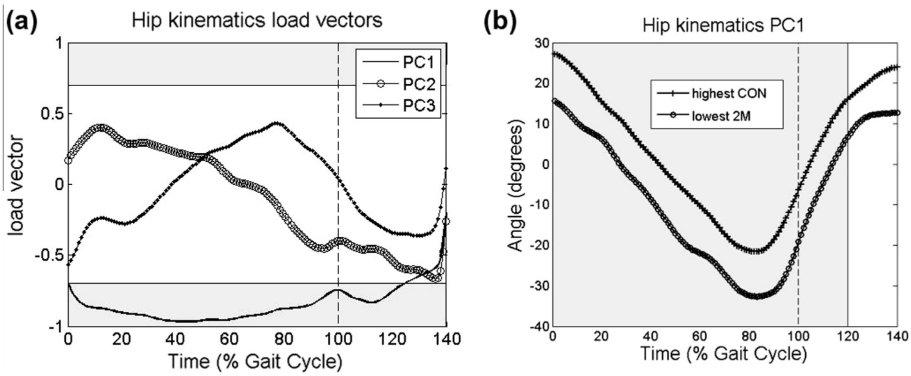

Fig. 6. Hip range of motion in sagittal plane: (a) load vectors for PC1, PC2 and PC3; (b) highest and lowest scores in 95% CI. Positive: hip flexion. Negative: hip extension. The grey area highlights the 0.71 threshold.

4. Discussion

The purposes of this study were to apply PCA analysis in joint net moment and range of motion waveforms to characterize the influence of different wedges on gait pattern of healthy elderly participants. Furthermore, some discrete parameters representative of net joint moment and kinematics were calculated in order to verify differences in the capacity of discriminating gait changes between PCA and the traditional approach. Both hypotheses established in the current study were accomplished: We identified the acute influence of wedges on the lower limbs' kinematics and net joint moment waveforms using PCA; and the PCA approach was more successful to determine changes in gait pattern compared to the traditional approach of discrete parameters. The results of the present study corroborates with a previous study (Deluzio & Astephen, 2007), which showed that PCA is a powerful method to analyze and to identify differences in gait data. Among the advantages observed with PCA analysis are: (a) it reduces the number of variables necessary to represent the whole gait waveform; (b) data from the entire gait cycle are considered; (c) data reduction results in a set of uncorrelated features that explain most of the variance presented in the data and; (d) significant differences that are not found in the traditional approach were found in PC scores comparisons.

The objective in analyzing the waveforms in the light of PCs is first to reduce the number of variables to analyze and then to find a biomechanical meaning for the PCs (Sadeghi et al., 2002a). The only variable that showed differences from CON in the discrete parameter (traditional approach) was $Ankle_{ROM}$ (Table 1), where the 2L conditions decreased the range of motion in the ankle joint. The choice of discrete parameters is often subjective, and the selected discrete parameters can be highly correlated (Olney et al., 1998). It is also often difficult to subjectively choose parameters that can adequately characterize the curves, and potentially meaningful parameters can easily be overlooked in subjective parameter extraction (Deluzio & Astephen, 2007). Besides, the difference in $Ankle_{ROM}$ observed with the discrete parameters was also noted in PC scores (Table 2).

Considering ankle kinematics, the lateral wedge conditions (1L and 2L) were those which showed more differences with statistical significance throughout the waveform compared to the CON condition (Fig. 4). During the intermediate stance phase, both lateral wedges decreased the ankle dorsiflexion, while at the end of the stance phase and beginning of the swing phase, the plantar flexion angles increased. This increased in dorsiflexion position previously to the toe off phase may cause a higher contraction of triceps surae muscles by myotatic reflex at toe off and, consequently, to promote a higher plantar flexion. Lee, Roan, and Smith (2009) found that ankle kinematics waveforms were also able to discriminate groups (healthy and obese participants).

Due to its application in patients with knee osteoarthritis, the influence of wedges on knee joint net moment in the frontal plane (knee valgus moment) was already investigated (Erhart et al., 2008; Franz et al., 2008; Kakihana et al., 2005; Kuroyanagi et al., 2007; MacLean et al., 2006; Schmalz et al., 2006). Some of them observed differences in knee valgus moment when wearing lateral wedges (Erhart et al., 2008; Kuroyanagi et al., 2007) and medial wedges (Franz et al., 2008; Schmalz et al., 2006); while no differences were observed by other authors (Kakihana et al., 2005; MacLean et al., 2006). All aforementioned authors analyzed discrete parameters. In the present study, the traditional approach indicated no differences between the wedges and CON. On the other hand, PC1 of knee joint net moment in the frontal plane explained the total foot contact during the stance phase and showed the 2L, 1P and 2P wedges as those able to reduce the valgus moment (Fig. 3b and Table 2).

With any of the techniques (PCA and traditional) differences were found in knee joint moments in the sagittal plane between the conditions and CON. In contrast, Deluzio, Wyss, Costigan, Sorbie, and Zee (1999) using PCA and comparing knee moments of healthy subjects and patients submitted to unicompartmental arthroplasty, observed that all three knee moments were able to differentiate the groups. These differences between the current and the cited study might be due to the fact that our participants were all healthy.

In summary, the lateral wedges presented differences between the wedge conditions and CON in all of the waveforms, except in the knee joint net moment in the sagittal plane (Table 2). This is consistent with other studies (Erhart et al., 2008; Kerrigan et al., 2002; MacLean et al., 2006; Schmalz et al., 2006) that used the traditional approach. Moreover, posterior wedges seemed to reduce the knee joint net moment in the frontal plane; whereas the medial wedges influenced the ankle joint. Overall, our findings suggest the kinetic and kinematic gait pattern in a healthy elderly population can be changed by wearing wedges.

Some limitations should be considered while assessing the findings of the present study. The joint moment data were normalized only by the participants' body weight (Winter, 1991), whilst the participants' body height was not considered. However, a previous study (Moisio, Sumner, Shott, & Hurwitz, 2003) showed that the body weight accounts for the most of the variance between subjects, and adding the body height only further 4%, 3% and 3% of the variance in joint moment data from the knee in the sagittal plane, knee in the frontal plane, and ankle in the sagittal plane would be explained, respectively (Moisio et al., 2003) and then we believe our normalization method was proper. Another important issue is that, according to Molenaar, Wang, and Newell (2013), the selection of 3 PCs could be underestimating the variance explained presented in the time domain analysis. However, in the current study the PCA model with 3 PCs accounted for about 90% of the variance explained in all assessed outcomes, and the whole waveform was evaluated. Thus, we believe the approach adopted reflected most differences presented in the assessed gait parameters. In addition, we only assessed the acute effect of wedges, where the gait pattern was verified after a short period of adaptation. Then we cannot warranty the maintenance of this gait pattern at middle and long-term. Finally, we only assessed healthy participants. Even elderly population being more susceptible to diseases related to bone degeneration, and to fall (Kirkwood, de Souza Moreira, et al., 2011), we cannot infer that the observed changes in gait pattern from our participants would be reproduced in an elderly population with pathological gait alterations. More studies should be carried out to analyze the influence of the wedges after a long-term adaptation and to determine if the results obtained in a healthy population are similar to those of a special one.

5. Conclusion

The PCA was successful to determine the influence of wedges on kinetics and kinematics aspects of gait in a healthy elderly population during gait. The influence of the wedges in the gait waveforms was evident using PCA, while with the traditional approach almost no differences among the conditions and CON were observed, which suggests that possible acute effects of wedges in gait could be hidden when discrete parameters are used. PCA analysis seemed to be a powerful tool to discriminate gait pattern among different wedge interventions considering the whole waveform of the analyzed parameter. The results obtained in a healthy elderly population might be useful to understand the change in kinematics and joint net moment behavior during gait and to provide quantitative data supporting wedge prescription for special populations.

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