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Prediction of mild anatomical leg length discrepancy based on gait kinematics and linear regression model

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ABSTRACT

Background: Leg length discrepancy (LLD) can be related to different pathologies, due to an inadequate distribution of mechanical loads, as well as gait kinematics asymmetries resulted from LLD.

Research question: To validate a model to predict anatomical LLD (ALLD) based on gait kinematics.

Methods: Gait of 39 participants with different lower limb pathologies and mild discrepancy were collected. Pelvic, hip, knee and ankle kinematics were measured with a 3D motion analysis system and ALLD, femur discrepancy (FD) and tibia discrepancy (TD) were measured by a computerized digital radiograph. Three multiple linear regression models were used to identify the ability of kinematic variables to predict ALLD (model 1), FD (model 2) and TD (model 3).

Results: Difference between peak knee and hip flexion of the long and short lower limb was selected by models 1 ($p < 0.001$) and 2 ($p < 0.001$). Hip adduction was selected as a predictor only by model 1 ($p = 0.05$). Peak pelvic obliquity and ankle dorsiflexion were not selected by any model and model 3 did not retain any dependent variable ($p > 0.05$). Regression models predicted mild ALLD with moderate accuracy based on hip and knee kinematics during gait, but not ankle strategies. Excessive hip flexion of the longer limb possibly occurs to reduce the limb to equalize the LLD, and discrepancies of the femur and tibia affects gait cycle in a different way.

Significance: This study showed that kinematic variables during gait could be used as a screening tool to identify patients with ALLD, reducing unnecessary x-ray exposure and assisting rehabilitation programs.

1. Introduction

Leg length discrepancy (LLD) is present in 4% to 90% of the population in the world depending on the criterion of clinical significance adopted by researchers [1,2]. It can be classified as anatomical (ALLD), when the difference between limbs can be directly measured on tibias, femurs or both, or functional discrepancies (FLLD), identified on postural analysis such as an excessive pelvic obliquity, excessive knee flexion or excessive ankle eversion on a standing position [3]. Both ALLD and FLLD have been related to different pathologies, such as knee and hip osteoarthritis, due to an inadequate distribution of mechanical loads [4,5], as well as gait kinematics asymmetries resulted from LLD have been related to plantar fasciitis [6], low back pain [7], and anterior knee pain [8].

The length of the lower limb can be measured by multiple methods. The most popular in clinical practice is the tape-measure technique, consisting of the distance between anterior superior iliac spine (ASIS) and the tip of the medial malleolus of each limb while lying in a supine position, and calculating the absolute difference between both, and radiographic studies are considered the gold standards for measuring leg length, but they expose patients to ionizing radiation [9].

Asymmetries in human gait kinematics have been associated to ALLD magnitude [4]. Pelvic elevation and hip adduction of the longer leg in single limb support phase of gait are kinematic features found in patients with different ALLD magnitudes [10,11]. Alterations of flexion of the hip, knee and ankle on the sagittal plane were found in some studies [10,12], but not in others [13,14]. The inconsistency of results may be related to the different level of leg discrepancy among studies

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and the presence of real or simulated discrepancy.

So, the identification of the kinematic features able to predict mild ALLD in subjects with lower limb pain would help health care providers to minimize the x-ray exposure and to improve rehabilitation of individuals by addressing both the anatomic discrepancy and the functional disorders resulted from ALLD. Biomechanical gait analysis is a valid and reliable tool to estimate tissue overloads and is already used to guide the selection of several treatment approaches in many clinical settings [15]. Thus, this study aimed at validating a multiple linear regression model to predict ALLD based on lower limb kinematics data during gait. The results of this study would contribute to increase the applicability of gait analysis in orthopedic facilities.

2. Methods

2.1. Participants

Sample size was determined as the number of participants necessary to reach a statistical power of 80%, with a coefficient of determination of 0.20 between variables and a moderate effect size and an $\alpha = 0.05$, using a bivariate normal model [16]. Thirty-nine subjects (21 females) with average age, mass and height of 43.0 ± 22.1 years, 71.2 ± 18.3 kg, 169.2 ± 11.8 cm, respectively, participated in the study. The inclusion criteria were subjects with previous history of injuries such as knee and hip osteoarthritis, back pain, heel pain and anterior knee pain (Table 1) with LLD magnitude ranging from 0 to 2 cm (Table 2). This group profile was chosen due to the correlation of these conditions to ALLD [4–8] and the range of ALLD was chosen because 90% of the population has some leg-length asymmetry [17] and the prevalence of LLD smaller than 2 cm has been reported to be higher than 99% of the population [18]. ALLD was measured by the computerized digital radiographs method as described in [19]. All radiographic exams were performed in the same laboratory and reviewed by the same Radiologists. The exclusion criteria were complaint of discomfort during x-rays or during biomechanical gait analysis. The study was approved by the local institutional Ethical Committee for Human Experiments. All participants signed an informed consent before participation.

2.2. Procedures

Initially, a standing trial in a static position was collected for each subject to individualize marker position, calculate joint centers and segment positions during walking. Then, participants performed a barefoot walk along an eight meters long walkway. Subjects were instructed to walk at their self-selected speed performing six trials along the walkway, and the last three gait cycles for each lower limb captured by the motion analysis system were used for analysis.

2.3. Data reduction

Kinematic data were collected using an eight high-speed cameras motion analysis system (Vicon, Oxford, UK) with a sample rate of 100 Hz. Markers were set according to Plug-In Gait recommendations [20]. Data were filtered by a second order Butterworth low pass filter, applied in the direct and reverse directions to avoid phase shift, with a cut-off frequency of 6 Hz, and Euler angles of lower limbs were calculated using Nexus software (Vicon, Oxford, UK), according to Grood and Suntay [21] proposal. To determine stance and swing phases of each cycle, the Foot Velocity Algorithm was used [22].

Predictors variables were the average of the three gait cycles of the peak absolute difference between the long and short leg of pelvic obliquity, hip flexion, hip adduction, knee flexion, ankle dorsiflexion and ankle eversion during stance phase (Fig. 1). Independent variables were ALLD, femur discrepancy (FD) and tibia discrepancy (TD), because some kinematics asymmetries occur due to the discrepancy of the tibia

and not the femur [10].

2.4. Data analysis

Pearson correlation coefficients were calculated to determine associations between peak of pelvic obliquity, hip flexion and adduction, knee flexion and ankle dorsiflexion and eversion during stance phase of walking and ALLD, FD, TD. Multiple linear regressions were developed for each independent variable. Stepwise approach was used to find the best model among all predictor variables possibilities, using the Akaike information criterion (AIC) to include variables into the models. Three independent regression models were developed using as independent variables ALLD (Model 1), FD (Model 2) and TD (Model 3). To assess the fitting of the model, a leave-one-out cross-validation procedure was used. All model coefficients were calculated using data from 38 subjects, and data from the subject left out of the analysis were used to predict ALLD. At the end, the mean absolute error (MAE) and the normalized MAE (%MAE) were used to assess the validation of the models:

$$\text{MAE} = \frac{1}{N} \sum_{i=1}^N |\hat{y}_i - y_i| \quad (1)$$

$$\% \text{MAE} = \frac{\text{MAE}}{\text{range(ALLD)}} * 100 \quad (2)$$

where \hat{y} represents simulated ALLD, y corresponds to the ALLD obtained through scanograms, and N is the number of subjects.

Statistical significance was set at $\alpha = 0.05$. All statistical analyses were performed using MATLAB (version 8.6.0, The Mathworks, USA).

3. Results

ALLD of the participants ranged from 0 to 1.96 cm, consisting of subjects classified with mild discrepancy (Table 1) and showed an average gait speed of 1.00 m/s (SD 0.27). Peak absolute differences of the predictor variables between the longer and shorter leg during stance phase of gait varied from 1.3° to 24.2° (Table 3). Correlations between hip adduction, and flexion and ALLD were significantly positive ($p < 0.05$, Table 4). None of the other correlations showed significant results for ALLD (Table 4). However, stepwise approach of linear regression model retained hip adduction, hip flexion and knee flexion in the model 1 (Table 5, $p = 0.007$, $r^2 = 0.29$, MAE = 0.35 cm, % MAE = 17.9%). Pearson partial correlation showed knee flexion to be an important controlling variable to hip flexion, increasing the correlation coefficient between hip flexion and ALLD to 0.45 ($p = 0.004$). Hip adduction was not affected by controlling knee flexion, maintaining correlation coefficient similar to the previous analysis.

Hip flexion presented significant correlation with FD (Table 4). Stepwise approach of linear regression retained both hip and knee flexion in model 2 (Table 5, $p < 0.001$, $r^2 = 0.33$, MAE = 0.29 cm, % MAE = 13.5%). There were no significant correlation between dependent variables and TD (Table 4). Pearson partial correlation showed knee flexion to be an important controlling variable to hip flexion, increasing the correlation coefficient between hip flexion and FD to 0.58 ($p < 0.001$). Model 3 did not retain any dependent variable ($p > 0.05$).

4. Discussion

This study sought to validate multiple linear regression models using lower limbs angles during gait to predict ALLD in subjects with lower limb injuries with different LLD magnitude (between 0 to 2 cm). According the revised literature, this is the first study that attempted to validate a regression model with that purpose and the results show a moderate but significant relation between kinematics variables and

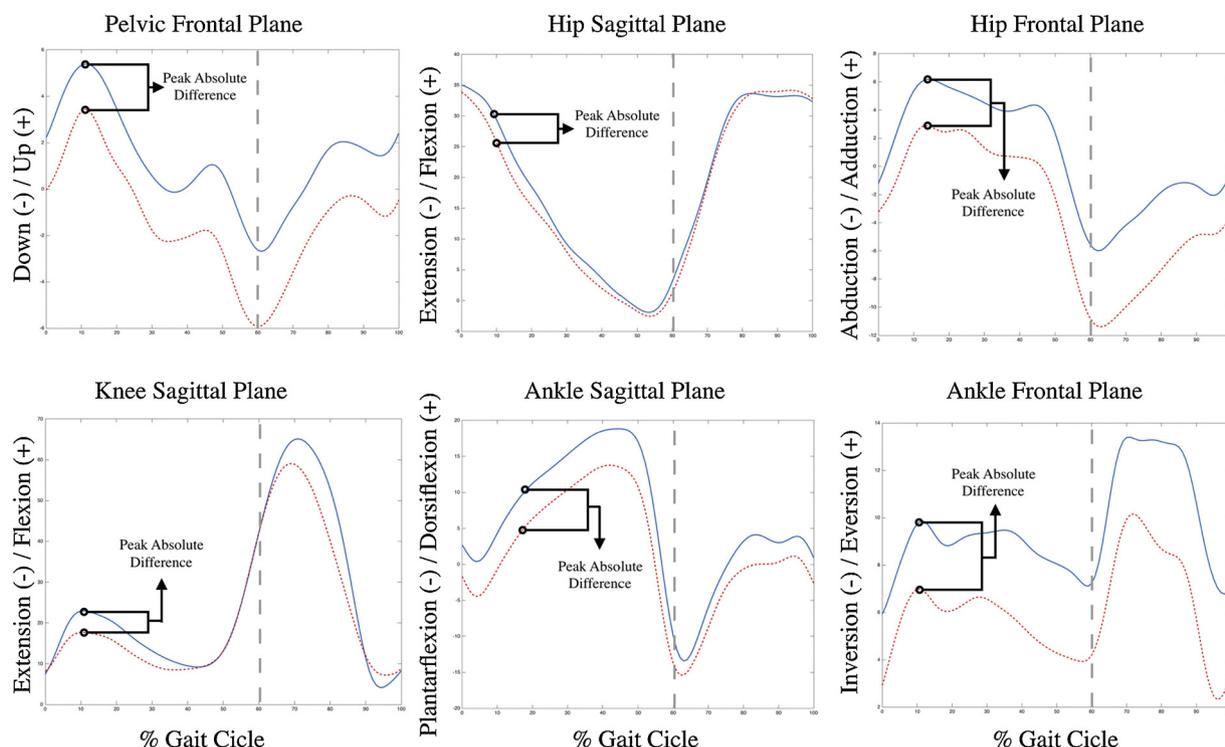


Fig. 1. The graphs above show angular behavior of the pelvis in frontal plane, hip in sagittal and frontal planes, knee in sagittal plane and ankle in sagittal and frontal planes. Solid curves represent long limb and dotted curves represent short limb. Vertical dotted line identifies the end of the stance and beginning of the balance. Peak absolute difference is the absolute difference between peaks of the lower limbs.

Table 1
Mean (Standard Deviation) of Femur, Tibia and Total Leg Length Discrepancy and Pain Level.

	Mean (SD)	Minimum	Maximum
Femur Discrepancy	0.50 (0.46)	0.00	2.15
Tibia Discrepancy	0.40 (0.37)	0.00	1.93
Total Discrepancy	0.62 (0.52)	0.00	1.96
Pain Level (VAS)	2.77 (2.08)	0.00	8.20

VAS: Visual Analogic Scale.

Table 2
Description of subjects' orthopedic disorders.

	N (%)
Low Back Pain	15 (38.5)
Anterior Knee Pain	8 (20.5)
Ankle Pain	6 (15.5)
Hip Pain	6 (15.5)
Plantar Fasciitis	2 (5)
Knee Osteoarthritis	2 (5)

Table 3
Mean (standard deviation) of the peak absolute difference angle between the longer and shorter leg during stance phase of pelvic obliquity, hip flexion, hip adduction, knee flexion, ankle dorsiflexion and ankle eversion and anatomic leg-length discrepancy.

	Mean (SD)
Pelvic Obliquity (°)	4.07 (3.13)
Hip Flexion (°)	2.23 (1.70)
Hip Adduction (°)	5.16 (4.63)
Knee Flexion (°)	3.76 (3.23)
Ankle Dorsiflexion (°)	2.42 (1.62)
Ankle Eversion (°)	1.90 (1.61)
Anatomic Leg-Length Discrepancy (cm)	0.62 (0.52)

Table 4
Pearson correlation coefficients between dependent and independent variables.

	ALLD	FD	TD
PO	0.28	0.19	-0.16
HF	0.35*	0.39*	0.09
HA	0.35*	0.22	-0.03
KF	0.08	0.02	-0.05
AD	0.26	0.28	-0.01
AE	0.18	0.06	0.01

ALLD: Anatomic Leg Length Discrepancy; FD: Femur Discrepancy; TD: Tibia Discrepancy; PO: Pelvic Obliquity; HF: Hip Flexion; HA: Hip Adduction; KF: Knee Flexion; AD: Ankle Dorsiflexion; AE: Ankle Eversion.

* p < 0.05.

asymmetries on lower limb length. Model 1 explained 29% of ALLD discrepancy with a MAE of 0.35 cm, while Model 2 explained 33% of FD discrepancy with a MAE of 0.29 cm, with an error ranging from 13% to 18% in both models. The error found in the present study is lower or similar to previous studies using tape measure to estimate ALLD [23,24].

Peak absolute difference of hip adduction between legs showed a positive correlation with ALLD (Table 4) and was retained in model 1 (Table 5). This result confirms previous studies proposing a moderate relation between hip adduction during human gait and ALLD and could be related to the pelvic obliquity down on the short limb and pelvic obliquity up on the long limb, commonly observed in patients with ALLD [10,13]. On a review of the literature Desai et al. [25] found that ALLD after primary THA may range from 1% to 27%. It has been associated with back pain, gait disorders, early loosening of components, gross dissatisfaction, morbidity, and need for revision surgery. Based on presented models, it is possible to differentiate if hip adduction asymmetry during gait is result of ALLD or not, providing helpful information to objectively address rehabilitation or mechanical axis compensatory measures of patients after THA.

Table 5

Coefficients (\pm standard error) of the linear models after stepwise approach (Only model 1 and 2 are presented, because model 3 did not retain any dependent variable).

	Constant	HF	HA	KF
Model 1 (ALLD)	0.0589 \pm 0.2245	0.1481 \pm 0.0524	0.0355 \pm 0.0179	-0.0624 \pm 0.0283
Model 2 (FD)	0.0812 \pm 0.1829	0.1851 \pm 0.0438	-	-0.0747 \pm 0.0238

ALLD: Anatomic Leg Length Discrepancy; FD: Femur Discrepancy; HF: Hip Flexion; HA: Hip Adduction; KF: Knee Flexion.

Peak hip flexion during stance phase was a predictor variable correlated to ALLD and, along with peak knee flexion, was included in model 1 (Table 5). These results corroborate with findings by Resende et al. [13] on healthy subjects gait with a simulated LLD with different height shoes, when shorter limbs had lower hip, knee and ankle flexion during stance when compared to the longer limbs. Also, Aiona et al. [10] found that children with ALLD had excessive knee flexion in the longer limb compared to the shorter one. Excessive hip and knee flexion may be an attempt to equalize center of mass by shortening the longer limb on a mild LLD, as evidenced by Song et al. [15] on LLDs greater than 2 cm.

Hip flexion was also selected by stepwise approach to get in model 2 (using FD as independent variable). However, in the model using TD as interest variable no kinematics variable was selected suggesting asymmetries on femur's length probably affects gait kinematics differently than tibia's length. Anatomical leg-length discrepancy could be due to a discrepancy either on the length of the femur or tibia, or both. Again AIONA et al. [10] have demonstrated that those types of ALLD can cause asymmetries in gait kinematics in different joints and segments of healthy children, partially corroborating the findings of the present study.

Stepwise approach selected peak knee flexion in both models (using ALLD and FD as dependent variable) and Resende et al. [13] found excessive knee flexion on the longer limb compared to shorter one. However, in resulting models the knee flexion coefficients were negative showing that the greater the ALLD, the smaller will be the knee flexion discrepancy between limbs. Participants of the present study showed an average gait speed (100 m/s), lower than those of Resende et al. [13] (145 m/s) and may explain different findings about knee flexion, since higher gait speed demands increased knee flexion contribution to dissipate loads during gait. It also should be considered the fact that the present study was performed on subjects with some sort of lower limb orthopedic conditions while Resende et al. [13] studied healthy subjects. So, the inclusion of the knee flexion in models probably might have occurred to correct model's prediction related to hip flexion, since partial correlation coefficient for this predictor variable increased when knee flexion was used as a controlling variable in both models 1 and 2.

Peak ankle dorsiflexion was not selected by stepwise approach as a predictor of any model, going against Resende et al. [13] findings. Unlike Resende et al. [13] we did not find ankle peak dorsiflexion on 'stepwise approach' as a predictor of any model. The reason could be accounted to the fact that while our interest variable was the real anatomic discrepancy theirs was over simulated ALLD. Also, as gait speed increases there is a trend to also increase plantar flexion moment of the short limb to control forward progression of the shank [13].

As limitations of this study we should consider that the statistical approach took into account a multivariate linear model, while kinematic variables of the human gait appear to have more complex interactions [26], so future research could employ non-linear models to predict ALLD. Also, subjects in the present study had different orthopedic conditions, which may affect gait symmetry in a particular manner. Additionally, the results should only be applied to subjects with mild discrepancy and similar gait speed. Future studies could be performed in order to evaluate the effect of different pathologies, higher gait speeds and greater ALLD on gait kinematics asymmetry.

In conclusion, this study finds that mild ALLD is related to exacerbated hip flexion and adduction during gait and multiple linear regression models allow predicting ALLD with moderate accuracy and that biomechanical gait analysis could be of great use as a screening tool to identify patients with ALLD and assisting rehabilitation programs. Femur discrepancy seems to have a greater influence on gait kinematics than the tibia one.

Conflict of interest

There are no conflict of interest.

Acknowledgments

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