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**Evaluation of spinal posture during gait with inertial measurement units**

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## **Abstract**

The increasingly number of postural disorders emphasizes the central role of the vertebral spine during gait. Indeed, clinicians need an accurate and non-invasive method to evaluate the effectiveness of a rehabilitation program on spinal kinematics. Accordingly, the aim of this work was the use of inertial sensors for the assessment of angles among vertebral segments during gait. Spine was partitioned into five segments and correspondingly five Inertial Measurement Units were positioned. Articulations between two adjacent spine segments were modeled with spherical joints and the tilt-twist method was adopted to evaluate flexion-extension, lateral bending and axial rotation. Eighteen young healthy subjects (9 males and 9 females) walked barefoot in three different conditions. The spinal posture during gait was efficiently evaluated considering the patterns of planar angles of each spine segment. Some statistically significant differences highlighted the influence of gender, speed and imposed cadence. The proposed methodology proved the usability of inertial sensors for the assessment of spinal posture and it is expected to efficiently point out trunk compensatory pattern during gait in a clinical context.

**Keywords:** spine biomechanics, posture monitoring, motion analysis, gait, multibody

## 1. Introduction

During human walking, the body is functionally divided into two units, passenger and locomotor. The first unit is composed by head, neck, trunk and arms, whereas the second one includes lower limbs and pelvis [1]. Although the names of these two units suggest that only the locomotor one contributes to the gait, also the upper body, and specifically the spine, plays an important role for the locomotion. In fact, the vertebral column carries out three essential tasks: the protection of the spinal cord, the transmittance of movement to the upper and lower extremities and the balance during both upright posture and ambulation [2].

Many common pathologies with postural disorders such as scoliosis, Parkinson's disease and low back pain often provoke an abnormal gait that negatively affects patients' quality of life [3–5]. Given the importance of rachis motion analysis, clinicians express the need for a method able to perform a segmental study of the spine with good accuracy and minimal invasion [6].

The multibody approach, typical of robotics, is increasingly used in biomechanics and suits well when the spine has to be modeled. This method defines a series of rigid segments connected by mechanical joints, which are identified with adequate degrees of freedom. The relative motion between two adjacent segments is assessed performing geometrical or motor assumptions on joints [7, 8]. The same modeling process can be applied on the human spine during a classical movement such as gait, by considering a serial linkage of spherical joints [9]. Ideally, in the vertebral column the number of rigid bodies is equal to the number of vertebrae; however, registering the kinematics of each vertebra is technically challenging. Therefore, usually subsets of adjacent vertebrae are considered as a single rigid link.

Among many different motion capture instrumentations, the most adopted for the spine evaluation during locomotion are the optical systems. In fact, they have the ability to capture complex and dynamic movements in a sophisticated way. Many literature works adopt optical motion capture systems to assess spine motion during gait. Syczewska et al. [3] focus on the correlation between gait pathology and degree of scoliotic deformity, while Ceccato et al. [10]

probed trunk muscles role in postural equilibrium. Characterization of physiological motion patterns of specific segments of the spine during gait are also investigated: trunk and shoulder in [11], lower thoracic and lumbar segment in [12]. Studies considering the whole spine and using a multi-segment model to assess kinematics are presented by Needham [13] and Leardini [14]. They consider 3 and 5 segments respectively and both estimate vertebral angular patterns of movement. As it is well known, despite their advantages, optical systems have also some limitations: high costs, markers occlusion, requirement of a lab setting and hence restricted volumes [15].

A more recent technological solution is represented by Inertial Measurement Units (IMUs). Their spread has begun with the rapid development of micro electromechanical systems. Wearable IMUs are lightweight, small-size, low power consumption, portable and low-cost [16]. Furthermore, they do not require laboratory constraints, allowing researchers to evaluate the movement in unobstructed environments [17]. Consequently, even if they entail a less dense spine segmentation with respect to markers, they respond to the need of physicians as a suitable technology for the clinical context.

IMUs are extensively adopted to assess gait spatio-temporal parameters from the motion of pelvis [18], shanks [19], feet [20] or multiple anatomical landmarks [21]. On the contrary, only few literature works use inertial sensors to evaluate spine posture. Some studies adopt IMUs for the evaluation of spine elementary and planar movements of flexion-extension, lateral bending and axial rotation obtained with the lower body in a static position [15, 22–26]. Other works use sensors placed on participants' back in order to estimate angles described during different clinical tests, such as performing the sit-to-stand transition, getting up and down from a step or assessing balance [27, 28]. In ergonomic studies, IMUs on the trunk are used to evaluate kinematics and posture during specific tasks: lifting objects from the floor and transfer objects from a spot to another [29, 30]. Moreover, some works use inertial sensors to evaluate spinal posture during gait, but they divide the column in few segments. More in detail, two works concentrate only on the pelvis, by comparing planar angles obtained through a single IMU to those estimated with markers of a

stereophotogrammetric system [27, 31]. Instead, Cafolla et al. [24] assess shoulders, trunk and pelvis angles with IMUs during gait, but without considering relative angles between adjacent segments.

A previous pilot study compared vertebral angular patterns measured with both optical and inertial systems, highlighting similar trends [32]. In light of these results, the aim of the present work was a more detailed assessment of spinal posture during gait with IMUs, estimating relative angles between adjacent vertebral segments during gait. Moreover, the influence exerted on ROM by the gender of the participants, the speed of walking and the prunitesence of an imposed cadence was evaluated.

## 2. Materials and Methods

### 2.1. Tilt-twist method

The vertebral column was modeled as a sequence of rigid segments connected by joints. The adopted segmentation was defined according to previous literature works using optical systems [3, 10–14] and five segments were identified: Cervical (C7-T6), Thoracic (T6-T12), Medium (T12-L3), Lumbar (L3-S1) and Sacral (S1).

Subsequently, the tilt-twist method was adopted [33, 34]. In the present study, according to this technique, the articulation between two adjacent spine segments was modeled by a spherical joint and the two bodies connected can be represented as two cylinders (Figure 1). The tilt-twist method allows to express the orientation of the frame fixed to a superior vertebra (units vectors  $\mathbf{i}_s, \mathbf{j}_s, \mathbf{k}_s$ ) with respect to the inferior vertebra frame (unit vectors  $\mathbf{i}_i, \mathbf{j}_i, \mathbf{k}_i$ ). Assuming the set of vectors  $\mathbf{i}', \mathbf{j}', \mathbf{k}'$  parallel to the set  $\mathbf{i}_i, \mathbf{j}_i, \mathbf{k}_i$ , considering the projection of the superior longitudinal axis with unit vector  $\mathbf{i}_s$  on the plane formed by unit vectors  $\mathbf{j}'$  and  $\mathbf{k}'$ , three angles between the two adjacent vertebral segments can be estimated:

- The tilt angle  $\varphi$ , developed with the bending movement of the superior segment with respect to the inferior one, measured between the two longitudinal axes [range:  $0^\circ; +180^\circ$ ];

- The tilt-azimuth angle  $\vartheta$ , pinpointing the plane formed by the longitudinal axes and defined by the bending movement [range:  $-180^\circ$ ;  $+180^\circ$ ];
- The twist angle  $\tau$ , developed with the rotation movement of the superior segment around its longitudinal axis [range:  $-180^\circ$ ;  $+180^\circ$ ].

From the combination of these three angles, planar movements of Flexion-Extension (FE), Lateral Bending (LB) and Axial Rotation (AR) can be evaluated [32].

\*\*\*\* Figure 1 near here \*\*\*\*

## 2.2. Participants

Nine males and nine females between 20 and 30 years old participated in the study giving their written informed consent. Subjects were chosen considering two exclusion criteria that could alter gait: (a) no musculoskeletal diseases in the previous three years and (b) no neurological disorders. Tables 1 shows mean and standard deviation values of participants' anthropometric data. This study was approved by the Local Institutional Review Board. All procedures were conformed to the Helsinki Declaration.

\*\*\*\* Table 1 near here \*\*\*\*

## 2.3. Instrumentation

Five Inertial Measurement Units (MTx, Xsens Technologies, Enschede, Netherlands) were adopted for the test. Each IMU consists of a tri-axial accelerometer, a tri-axial gyroscope and a tri-axial magnetometer. Accelerometers and gyroscopes measurement ranges were set to  $\pm 50 \text{ m/s}^2$  and  $\pm 1200 \text{ dps}$  respectively. The sampling frequency was fixed at 50 Hz. According to the previous segmentation of the vertebral column, IMUs were positioned on participants back at C7, T6, T12, L3 and S1 vertebral levels (Figure 2). In this way, each sensor was considered integral with the

segment underneath it (C7 for Cervical segment, T6 for Thoracic segment, T12 for Medium segment, L3 for Lumbar segment and S1 for Sacral segment).

The sensors on C7 and T6 vertebrae were fixed with adhesive tape. Elastic bands were adopted to fix sensors on T12, L3 and S1 levels. C7, T6 and S1 IMUs were positioned with x-axis pointing upward, z-axis opposite to walking direction and y-axis forming a Cartesian triad with the other two. To avoid sensors collision and misplacement, T12 and L3 sensors were rotated counter-clockwise by 90 degrees, obtaining the z-axis opposite to walking direction, the y-axis pointing downward and the x-axis forming a Cartesian triad with the other two. The software used for data acquisition was Xsens-MT Manager. Figure 2 shows IMUs configuration adopted in the study. In particular, the vertebral column segmentation and IMUs' local reference systems are depicted respectively in Figure 2A and in Figure 2B.

### *2.3. Protocol*

The experiment was conducted indoor. A 20 meter long straight path was marked on the floor by means of indicators. After wearing the five inertial sensors on the back as described above, all the participants were first asked to stand still for a few seconds; this was assumed as the neutral posture and the corresponding orientation of IMUs was acquired. Then, subjects were asked to walk barefoot along the path in 3 different conditions:

1. Normal speed with metronome. The cadence was set to 2.0 steps/s [35].
2. Slow speed with metronome. The cadence was set to 1.5 steps/s [35].
3. Self-selected comfortable speed with no metronome.

For each condition, participants walked along the path for three times always in the same direction. Before the beginning of tests with the metronome, subjects practiced the rhythm to easily synchronize heels impacts with the sound.

\*\*\*\* Figure 2 near here \*\*\*\*



#### 2.4. Signal Processing and Data Analysis

Custom Matlab® routines were developed to process signals and analyze data. The same analysis was conducted for all test conditions and for all participants. Since each IMU was considered integral with the underlying vertebral segment, five segments were defined (Cervical, Thoracic, Medium, Lumbar and Sacral) and their relative motion during gait was evaluated.

IMUs data rearrangement. The analysis concentrated on the output of each IMU, which was a matrix of  $n$  rows (where  $n$  was the number of samples of the recording) and 16 columns. The first column was the time count; columns from two to seven contained acceleration and angular velocity values referred to the three axes of the sensor; the last nine columns were the direction cosines of the IMU's reference frame with respect to the global reference system. For each sample, these nine values were rearranged in a 3x3 matrix  ${}^G R_S$  that expressed the global orientation of the sensor:

$$R_S^G = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \quad (1)$$

where lower-case S is the sensor, upper-case G is the global reference frame and each matrix element  $R_{ij}$  is the cosine of the angle between i-axis of S and j-axis of G.

With the subject in his/her own neutral posture, 100 recorded samples were considered to evaluate the reference orientation of each sensor, stored as matrix  $R_{S,0}^G$ . Subsequently, during walking sessions, the orientation of each sensor with respect to its reference orientation (matrix  $R_S^{S,0}$ ) was calculated for each sample.

In order to represent the angular motion of a segment with respect to the adjacent inferior one by means of the tilt-twist method, the orientation matrix  $R_{S,\text{sup}}^{S,\text{inf}}$  of the superior sensor S.sup with respect to the inferior sensor S.inf had to be evaluated:

$$R_{S,\text{sup}}^{S,\text{inf}} = \left( R_{S,\text{inf}}^{S,\text{inf},0} \right)^{-1} R_{S,\text{sup},0}^{S,\text{inf},0} R_{S,\text{sup}}^{S,\text{sup},0} \quad (2)$$

where:

$R_{S.inf}^{S.inf.0}$  and  $R_{S.sup}^{S.sup.0}$  are the orientation matrices of the inferior and superior sensors with respect to their reference pose respectively;

$R_{S.sup.0}^{S.inf.0}$  is the relative orientation matrix between the two sensors in the neutral posture, given by:

$$R_{S.sup.0}^{S.inf.0} = (R_{S.inf.0}^G)^{-1} R_{S.sup.0}^G \quad (3)$$

Gait Cycle identification. In order to correlate spinal posture with locomotion, gait events were identified and gait cycles were defined. Two signals from the T12 IMU were analyzed for this purpose: the acceleration along the anterior-posterior axis (z-axis) and the angular velocity around the vertical axis (y-axis). The first signal was adopted to select Heel-Strike (HS) and Toe-Off (TO) respectively as maximum and minimum peaks; then, the sign alternation of the second signal was evaluated to distinguish between right and left sides [36]. Considering the first and the last HS over the 20m-path, the mean cadence was evaluated. For the two conditions with the metronome, this value was used to verify if subjects correctly followed the rhythm. Table 2 shows the mean values of cadence adopted by the participants during the three conditions.

\*\*\*\* Table 2 near here \*\*\*\*

Tilt-twist method application. By applying the tilt-twist method, movements of FE, LB and AR were evaluated from the combination of tilt, tilt-azimuth and twist angles [32]. For both males and females and for the three walking conditions, the angular patterns of every vertebral segment with respect to the inferior adjacent one were calculated (Cervical-Thoracic, Thoracic-Medium, Medium-Lumbar, Lumbar-Sacral). The sacral segment was considered the beginning inferior and its angular pattern (Sacral) was referred to its initial neutral orientation.

All the patterns were normalized to the gait cycle, from 0% to 100%; subsequently, all gait cycles were mediated first intra-subject and then inter-subjects.

ROM evaluation. From the angular patterns obtained, ROM were estimated as differences between the maximum and the minimum angular values inside each gait cycle. Then, ROM were

averaged first intra-subject and then inter-subjects. The non-normal distribution of data verified with the Shapiro-Wilk test imposed a non-parametric statistical analysis, which was conducted to evaluate the influence of three aspects:

- The gender of participants. The Mann-Whitney U test (2-tails, significance level:  $\alpha = 0.05$ ) was performed between males ROM and females ROM for each couple of segments (5 couples) in each plane (3 planes) and for each testing condition (3 conditions). Consequently, Mann-Whitney U test was repeated 45 times to evaluate the influence of gender on ROM.
- The speed of walking. Since only few statistical differences were identified with the Mann-Whitney U test between males and females, the effect of walking speed on ROM was investigated by considering all the eighteen subjects together without gender distinction. In detail, the Wilcoxon signed-rank test (2-tails, significance level:  $\alpha = 0.05$ ) was performed between normal speed ROM and slow speed ROM for each couple of segments (5 couples) in each plane (3 planes). Consequently, the Wilcoxon signed-rank test was repeated 15 times to evaluate the influence of walking speed on ROM.
- The presence/absence of metronome. Since only few statistical differences were identified with the Mann-Whitney U test between males and females, the effect of the presence/absence of metronome on ROM was investigated by considering all the eighteen subjects together without gender distinction. In detail, the Wilcoxon signed-rank test (2-tails, significance level:  $\alpha = 0.05$ ) was performed between normal speed ROM and self-selected speed ROM for each couple of segments (5 couples) in each plane (3 planes). Consequently, the Wilcoxon signed-rank test was repeated 15 times to evaluate the influence of metronome on ROM.

### **3. Results**

For each walking condition, 36 gait cycles were selected and considered for the estimation of spine angular patterns and ROM. In all, 108 gait cycles were analyzed for each subject.

### 3.1. Angular patterns

Since angular patterns of males and females showed the same trend for the three conditions, only those of normal speed are reported as an example. Figures 3 and 4 show males and females angular patterns when walking at normal speed respectively. Every figure is divided into 15 plots. In every plot there is the mean value of angular pattern between two adjacent segments in a specific plane. Consequently, every row of three plots corresponds to a relative angle of two adjacent segments (Cervical = Cer, Thoracic = Tho, Medium = Med, Lumbar = Lum, Sacral = Sac) and every column of five plots corresponds to a planar movement (FE, LB and AR). Moreover, every plot contains the standard deviation band and three vertical lines. The first dashed line represents the left TO and occurs at approximately 10% of the cycle; the second continuous line identifies the left HS and occurs at about 50% of the cycle; finally, the right TO is represented by the third dashed line, which occurs at approximately 60% of the cycle [1].

### 3.2. Ranges of Motion

Figures from 5 to 9 contain bar diagrams for the comparison of mean ROM in different conditions. Every figure is divided into 3 subplots, each of which is related to a planar movement (FE, LB and AR). On every subplot there are 5 couples of columns, one for every pair of adjacent segments (C-T for Cervical-Thoracic, T-M for Thoracic-Medium, M-L for Medium-Lumbar, L-S for Lumbar-Sacral and S for Sacral). Each couple of columns is composed of a grey one and a white one, which are related to two different conditions as pointed in the legends. Furthermore, standard deviation values are reported with black lines on each column. When p-values obtained from the statistical test are  $\leq 0.05$ , an asterisk is reported above the couple of columns. More specifically, Figures from 5 to 7 refer to the gender influence on ROM at three different walking conditions: normal speed, slow speed and self-selected speed. Figure 8 shows the influence of walking speed on ROM, comparing the trials at normal and slow speed without gender distinction. Finally, Figure 9 shows the influence of metronome on ROM, comparing the trials at normal speed to those at self-selected speed without gender distinction.

\*\*\*\* Figure 3 near here \*\*\*\*

\*\*\*\* Figure 4 near here \*\*\*\*

\*\*\*\* Figure 5 near here \*\*\*\*

\*\*\*\* Figure 6 near here \*\*\*\*

\*\*\*\* Figure 7 near here \*\*\*\*

\*\*\*\* Figure 8 near here \*\*\*\*

\*\*\*\* Figure 9 near here \*\*\*\*

## **4. Discussion**

The aim of this study was the evaluation of spinal posture during gait by using inertial sensors. First, planar angular patterns described by vertebral segments with respect to the inferior adjacent ones were estimated with the tilt-twist method. Then, ROM of these movements were calculated. The discussion focuses on these two parts, but it also provides an overview of methodology, results and limitations of the study.

### *4.1. Evaluation of angular patterns in comparison with literature*

As can be seen in Figures 3 and 4, both genders in different conditions demonstrated similar patterns of motion for all segments and all planes.

Considering FE patterns, in according to [12], a two-phase movement occurs inside the gait cycle of every segments, corresponding to one flexion-extension cycle per step. In this case, the

trend is more evident for male subjects. In particular, for Sacral segment, this pattern is justified because the pelvis performs an extension at the first right heel-strike and then progressively reverses its tilt to the next left heel-strike; the same trend occurs in the second half of the cycle. Another aspect that can be highlighted concerns the patterns of Lumbar-Sacral and Sacral segments, which are complementary for both males and females in all the conditions [12].

Considering LB patterns, always in according to [12], peaks occur at approximately 15% and 65% of the gait cycle, coinciding with early swing phase. Furthermore, similarly to [12, 13], the displacement of lower thoracic and lumbar regions is towards the weight bearing limb, while the displacement of the pelvic segment is towards the swinging side. This aspect can be noticed comparing the sign of peaks occurring at 15% and 65% of the gait cycle: the first one is positive for the couples of Thoracic-Medium, Medium-Lumbar and Lumbar-Sacral and negative for Sacral; the second one follows the opposite trend. The pattern of Cervical-Thoracic couple has a behavior similar to the one of Sacral segment, as if the cervical area follows the pelvis inclination during the cycle.

Considering AR patterns, a similarity is identified with the pelvis pattern found by [31], taking into account the different axes convention. There is a negative peak coinciding with the first right HS, then a positive peak in correspondence of the left HS and finally another negative peak with the second right HS. This explains that, when a right stance occurs, the pelvis performs a rotation to the right side; on the contrary, during the left stance, the pelvis rotates to the left side. It is also possible to note that lower thoracic and lumbar segments have an opposite trend, whereas the upper thoracic region (Cervical-Thoracic) follows the behaviour of Sacral segment with a less consistent pattern.

#### *4.2. Statistical evaluation of ROM*

As can be seen in Figures from 5 to 9, mean values of ROM are in a range from 1° to 12°, which is consistent with previous findings in literature [13, 14, 37]. The same consideration can be done with respect to standard deviation values found by [37].

First, the influence of the gender on ROM values is considered, comparing males and females ROM in all planes and all conditions. Male subjects show greater ROM for the FE of Cervical-Thoracic segments in all conditions: normal speed with metronome ( $p=0.06$ ), slow speed with metronome ( $p=0.02^*$ ) and self-selected speed ( $p=0.03^*$ ). The difference due to the gender disagrees with some previous results in literature, according to which females have a greater cervical ROM with respect to males [38, 39]. This aspect could be investigated with further analysis, involving more subjects. Another gender difference is found for LB: females show greater lumbar ROM than males for all the three conditions. Even if no statistically significant difference is found, these results are in agreement to [40], according to which the greater lumbo-sacral motion of women has the aim to reduce vertical COM displacement and thus to conserve energy during walking.

Subsequently, the influence of walking speed is considered, comparing normal speed and slow speed conditions without distinguishing between males and females. For LB, the reduction of speed implies a reduction of ROM for all couples of segments ( $p=0.02^*$  for Thoracic-Medium and  $p=0.00^*$  for Sacral). This result is in agreement with [37], which affirms that LB segmental motion increases with the walking speed. For the other two planes the behavior of pelvis differs from that of other segments. In fact, for AR the speed reduction provokes a general ROM reduction ( $p=0.03^*$  for Medium-Lumbar,  $p=0.00^*$  for Lumbar-Sacral); on the contrary, with the change from normal to slow speed, Sacral segment increases its ROM ( $p=0.04^*$ ). For FE on the sagittal plane the trend is opposite, without significant differences: the reduction of speed increases ROM of all segments apart from the Sacral one. It seems that, for the pelvis, the speed reduction provoked an increase of AR ROM, at the expense of FE and LB ROM. The influence of walking speed on spine ROM could be better investigated by imposing two cadences that are more far apart.

Finally, the influence of the metronome was considered, comparing normal speed and self-selected speed conditions without distinguishing between males and females. Apart from the AR of Lumbar-Sacral segments ( $p=0.05^*$ ), no significant differences or similar trends were found in all

planes and for all couples of segments. This suggests that the presence or the absence of an external walking rhythm did not influence spinal ROM of participants.

#### *4.3. Overview of methodology and results*

The methodology adopted for the present study provided a protocol suitable to evaluate spinal posture during gait in a clinical set. In fact, many factors contributed to make this protocol robust and to reduce the time needed to acquire and analyze data: the positioning of IMUs on participants' back is repeatable; each user's neutral position is evaluated and taken into account for the estimation of angular pattern during gait; the assessment and testing of the spine ROMs in different gait conditions take advantages of a repetitive algorithm. Moreover, the methodology is expected to efficiently point out trunk compensatory pattern during gait and it is a worth indicator of clinical and rehabilitative effectiveness (e.g. Parkinson, Pisa syndrome, scoliosis, Trendelenburg gait).

#### *4.4. Limitations*

The first limitation was introduced considering vertebral segments as rigid links with the adoption of a multibody approach. Moreover, the absence of previous works evaluating relative angles between adjacent vertebral segments during gait with IMUs limited the discussions. In fact, results were necessarily compared with those from articles estimating the same values through optical motion capture systems. Another restriction was due to the different spine segmentations adopted by other literature studies; this made it more difficult to compare patterns of the present work with those of the previous ones. In addition, even if the union of males and females ROM for the evaluation of walking speed and metronome effects was justified, the application of non-parametric statistical tests on wide ranges of data may have influenced part of results. About this aspect, further investigations are planned to define more homogeneous groups of subjects and to increase the statistical significance of results.

### **5. Conclusions**

To the best knowledge of the authors, the present study was the first to use IMUs for the assessment of relative angles between vertebral adjacent segments during gait. The estimated



patterns were similar to those found in literature with optical motion capture systems. Also the ROM were in agreement with previous works, highlighting some significant differences between genders and walking speeds.

Overall, the results demonstrated the suitability of this technology for the clinical context, which requires low-cost, non-invasive and accurate methods. In fact, the possibility to use IMUs allows clinicians to assess the effectiveness of a rehabilitation program or to schedule a therapy with a direct evaluation of the patient at home. Future plans involve the realization of a database with angular patterns of healthy people of age different groups, in order to highlight differences and to provide a frame of reference. Finally, starting from observations obtained from this database, future studies could apply the same methodology to patients with Parkinson's disease, Pisa syndrome or scoliosis, in order to evaluate the same patterns and ROM in pathologic subjects.

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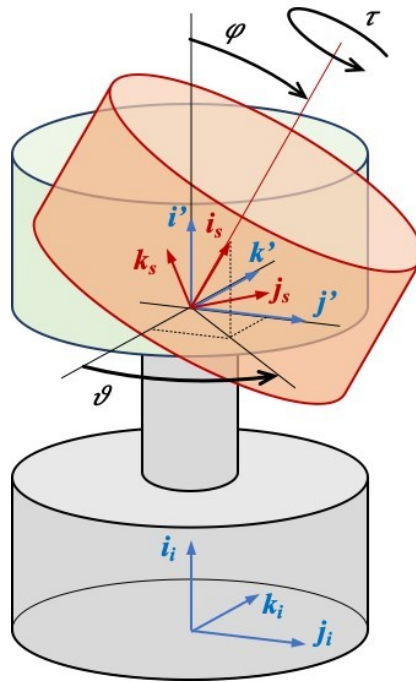
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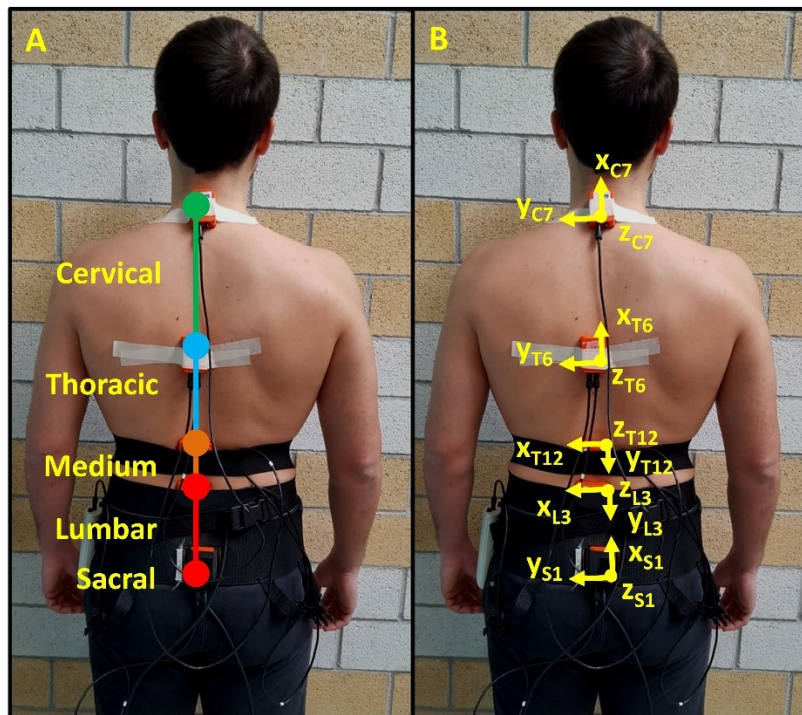
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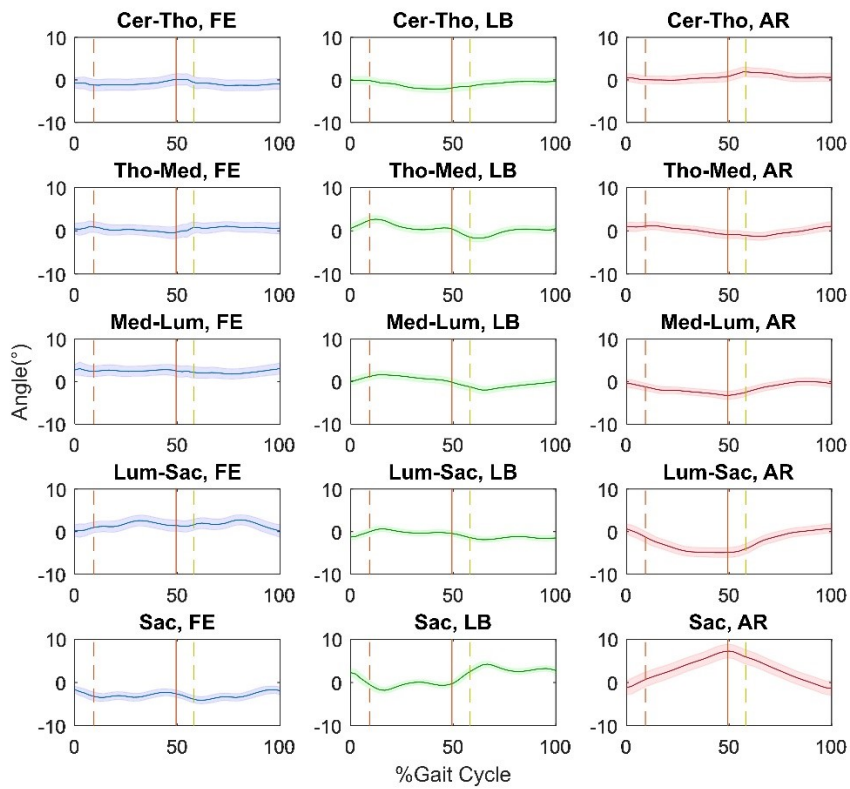
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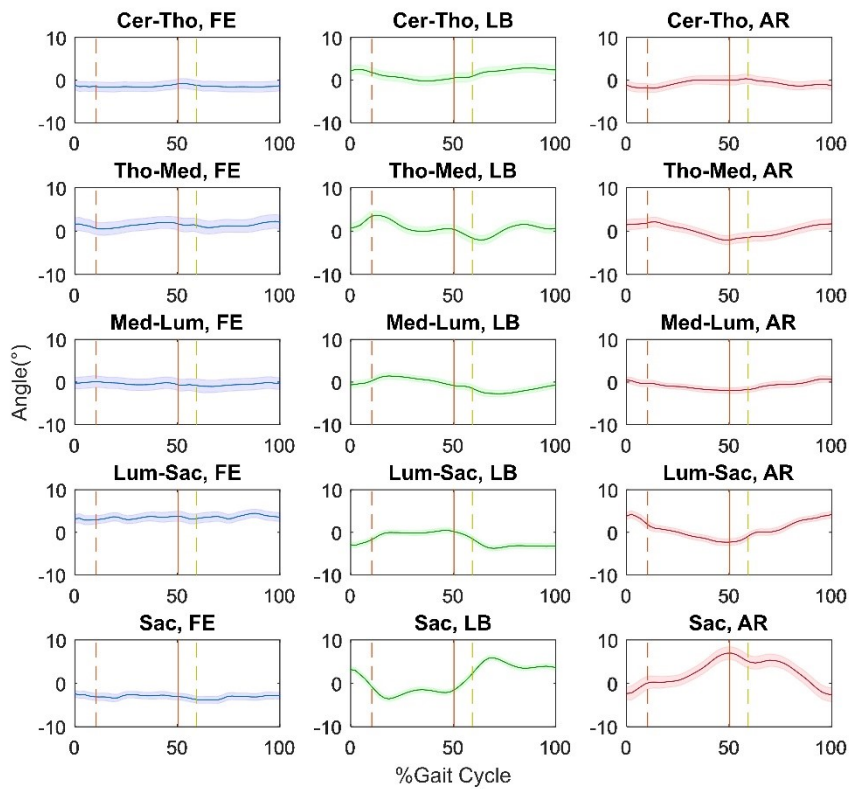
**Figure 1.** Tilt ( $\varphi$ ), tilt-azimuth ( $\theta$ ) and twist ( $\tau$ ) angles of the upper segment with respect to the lower one.



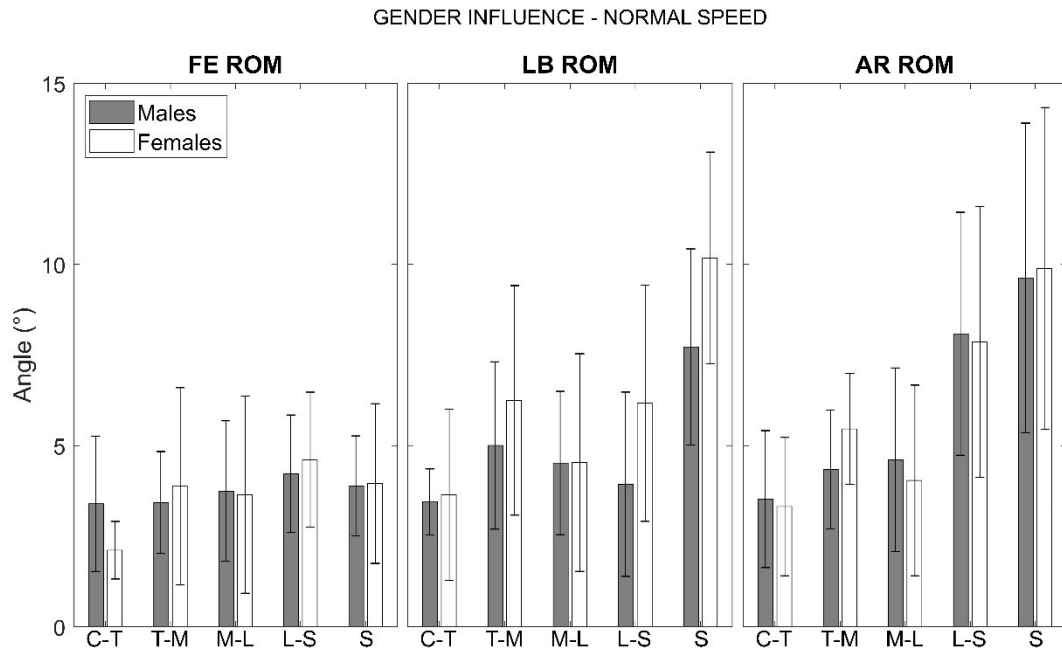
**Figure 2.** A) Vertebral column segmentation: Cervical (from C7 to T6), Thoracic (from T6 to T12), Medium (from T12 to L3), Lumbar (from L3 to S1) and Sacral (corresponding to S1). B) IMUs local reference systems.



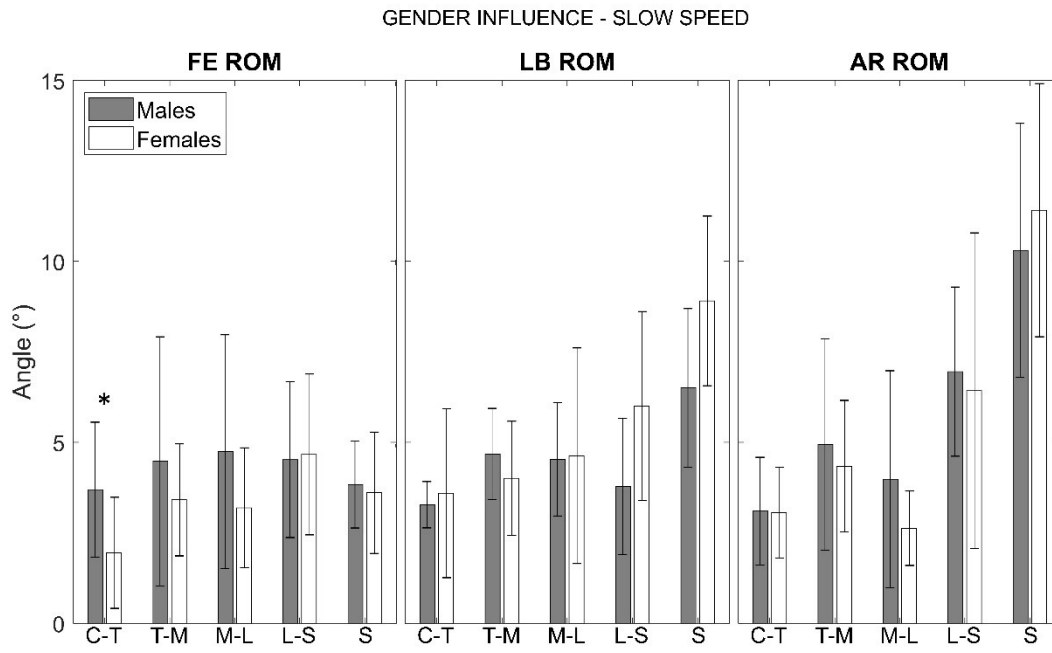
**Figure 3.** Males angular patterns for normal speed with imposed cadence.



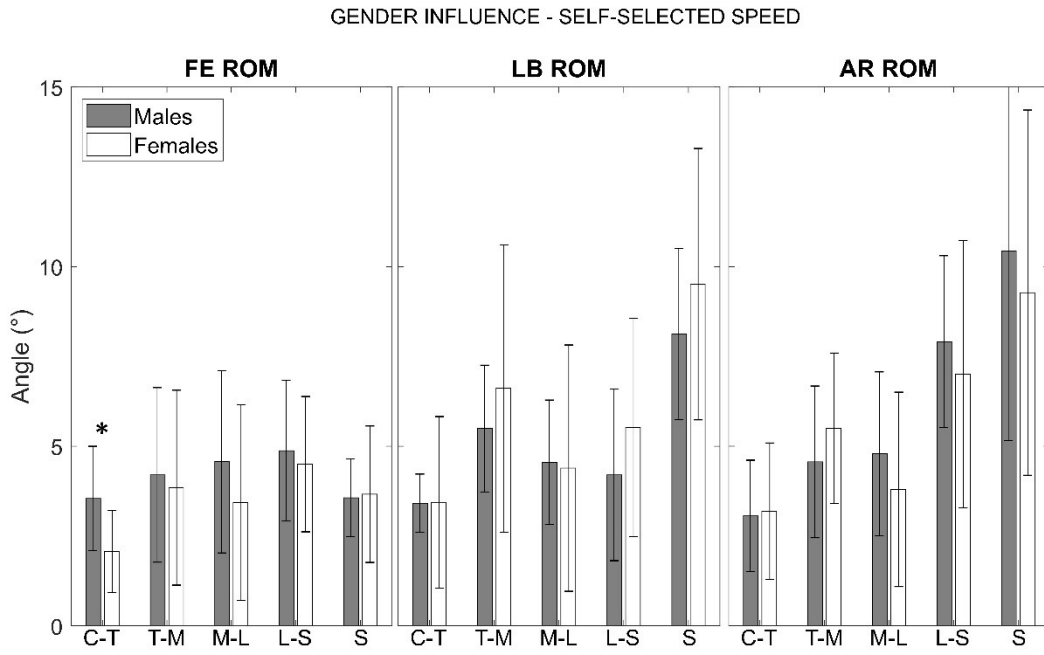
**Figure 4.** Females angular patterns for normal speed with imposed cadence.



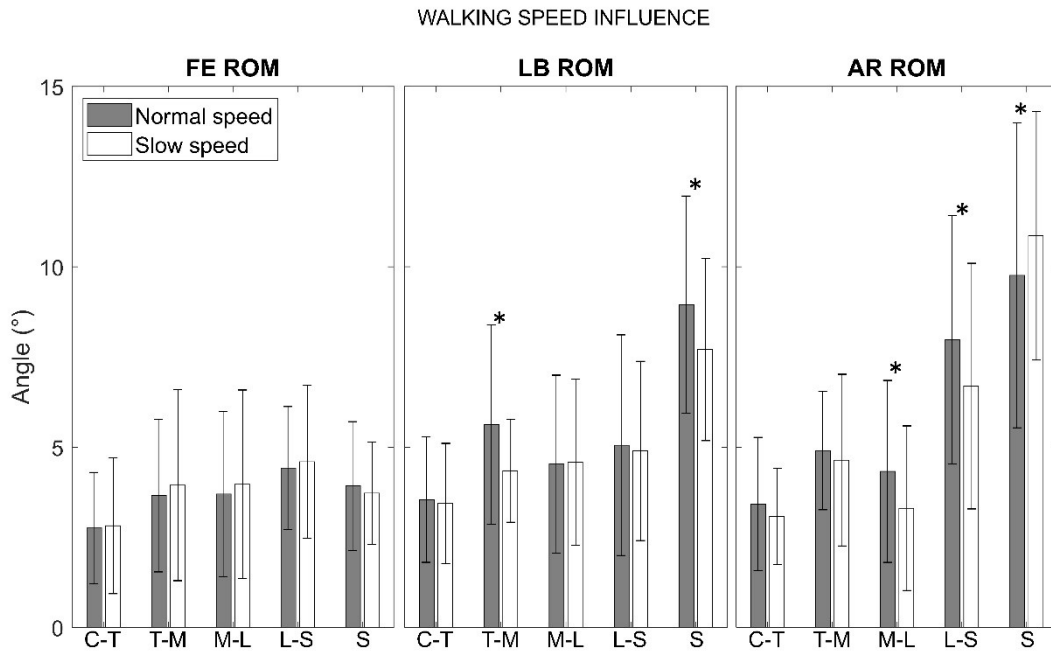
**Figure 5.** Influence of gender on ROM at normal speed.



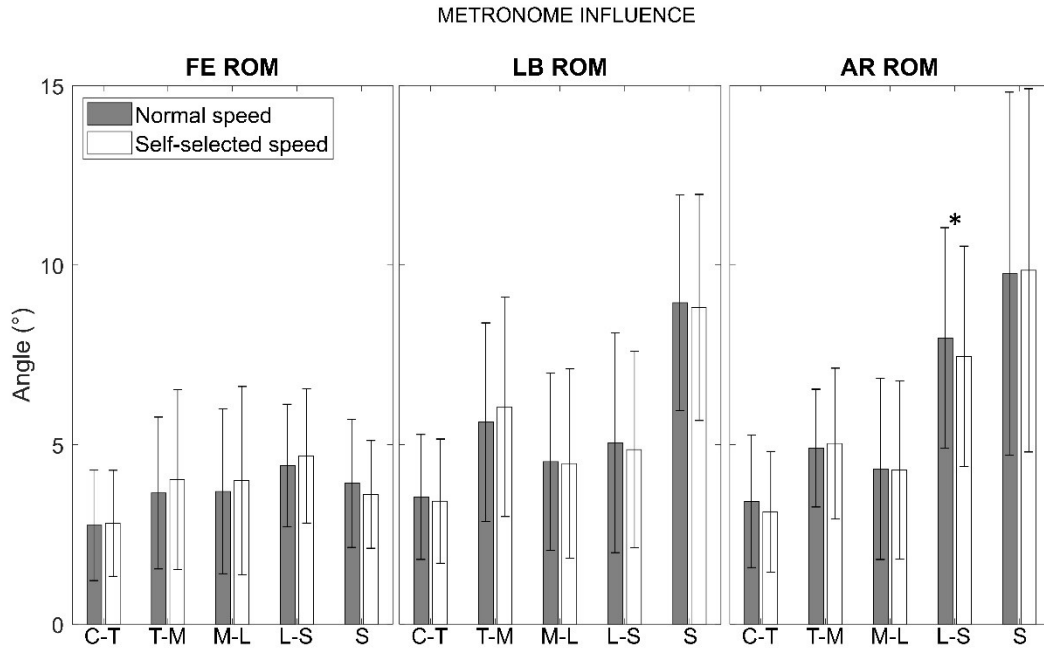
**Figure 6.** Influence of gender on ROM at slow speed.



**Figure 7.** Influence of gender on ROM at self-selected speed.



**Figure 8.** Influence of walking speed on ROM.



**Figure 9.** Influence of imposed cadence on ROM.

**Table 1.** Anthropometric data of participants (mean  $\pm$  standard deviation): Age (years); BMI ( $\text{kg}/\text{m}^2$ ); Ac = acromions distance (cm); Up = upper arm length (cm); Tr = trunk length (cm); Th = thigh length (cm); Sh = shank length (cm); Pu = pubis – ground distance (cm).

	Females	Males
Age (years)	26.3 $\pm$ 1.8	25.2 $\pm$ 0.7
BMI ( $\text{kg}/\text{m}^2$ )	20.7 $\pm$ 2.1	24.4 $\pm$ 1.6
Ac (cm)	37.6 $\pm$ 2.5	45.0 $\pm$ 3.3
Up (cm)	53.2 $\pm$ 4.9	59.7 $\pm$ 3.5
Tr (cm)	58.0 $\pm$ 3.0	67.2 $\pm$ 3.4
Th (cm)	46.1 $\pm$ 4.4	49.2 $\pm$ 4.3
Sh (cm)	45.4 $\pm$ 3.9	52.0 $\pm$ 2.1
Pu (cm)	74.2 $\pm$ 4.5	79.6 $\pm$ 4.3

**Table 2.** Values of walking cadence during the test for females and males (mean  $\pm$  standard deviation).

	Males	Females
Normal speed (metronome, 2.0 steps/s)	1.97 $\pm$ 0.03	1.98 $\pm$ 0.10
Slow speed (metronome, 1.5 steps/s)	1.45 $\pm$ 0.09	1.53 $\pm$ 0.04
Self-selected speed (no metronome, steps/s)	1.86 $\pm$ 0.07	1.81 $\pm$ 0.08