Original Article:

Analysis of gait data in manifold space: The control of the centre of mass during gait.

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Abstract

This study investigated the feasibility of the uncontrolled manifold approach (UCM) to analyse gait data variability in relation to the control of the centre of mass (COM) in adults with and without neuropathology. The proposed method was applied to six able-bodied subjects to characterise mechanisms of normal postural control during stance phase. This approach was repeated on an early stroke patient, who attended the laboratory three times at three monthly intervals, to characterise the variability of COM movement during walking with and without an orthosis. Both able-bodied subjects and the stroke participant controlled COM movement during stance but utilized a different combination of lower limb joint kinematics to ensure that the COM trajectory was not compromised. Interestingly, the stroke subject, despite a higher variability in joint kinematics, was able to maintain a stable COM position throughout stance phase. The stabilisation of the COM decreased when the patient walked unaided without the prescribed orthosis but increased over the six months of study. The UCM analysis demonstrated how a stroke patient used a range of lower limb motion pattern to control the COM. It is suggested that this analysis can be used to track changes in these movement patterns in response to rehabilitation. As such we propose that this approach could have clinical utility to evaluate and prescribe rehabilitation in stroke patients.

1. Introduction

Many stroke survivors present with an altered ability to walk. Compensatory actions and gait strategies are often adopted to achieve a safe walking activity. Motor control is a key issue for those people who have suffered from a stroke and for whom limited joint coordination impairs their mobility. An understanding of how the central nervous system (CNS) compensates to control motion following a stroke may inform subsequent therapy.

The theory of the uncontrolled manifold (UCM) has been recently introduced (Latash et al., 2007; Scholz and Schöner, 1999) to investigate how the CNS acts with respect to selected motor tasks by choosing combinations of different musculoskeletal elements that are involved in the performance of the task. That is to say CNS may employ a variety of different approaches to achieve a task. Exploiting this approach it may be possible to predict which motor variables the CNS controls and what are the elements/degree of freedoms (DOFs) that it has to organise for that particular motor task to be performed. This theory can thus be seen as an analysis of the variability of a selected functional task in a multi-degree of freedom system. The variability can either be "good", if the task goal remains unaltered, or "bad", if deviations from it occur. The UCM itself is a subspace of all possible combinations of motor elements (elemental variables) that lead to a consistent value of a performance variable. For example all the different combinations of lower limb joint angles that together place the centre of mass (COM) in a certain position in 3D space define a UCM subspace. It is defined "uncontrolled" because the control of the variability within it is unnecessary as all the combinations (i.e., set of lower limb joint angles) within that subspace preserve the performance variable value (i.e., 3D position of the COM) (Scholz and Schöner, 1999). Thereby, the UCM approach can also be seen as a method to quantify synergies. In this context, a synergy refers to an organization of elemental variables that stabilises a performance variable (Latash and Anson, 2006). A practical example could be to use the UCM approach to understand how the CNS organises joint angles (elemental variables) to allow a smooth COM movement (performance variable) and thus safe locomotion. It is important to mention that variability across trials is partitioned into two components: one that lies within the UCM and one that is perpendicular to the UCM. These two variabilities, expressed as indexes of variance across repetitions of the same task, are used to verify the hypothesis about the aspects of movement that are controlled. If the variance within the UCM is bigger than the one perpendicular to it, the hypothesis about the stabilisation of the selected motor task is accepted.

This analysis can provide clinicians with a better understanding of motor coordination and its relationship with rehabilitation approaches providing an explanation on how different dynamic resources can lead to a successful motor performance. Having information on the behaviour of the system will allow a more specific and

individualised treatment to accelerate recovery as the intervention target (musculoskeletal elements) can be identified. Which movement variations should be encouraged and which discouraged? An answer to this question will advance clinical practice and outcomes for stroke survivors.

The UCM analysis method has recently been used to verify the control of motor task predominately related to the upper extremity, sit-to-stand, standing and hopping performances of able-bodied and impaired subjects (Auyang et al., 2009; Domkin et al., 2002; Freitas et al., 2006; Hsu et al., 2007; Reisman and Scholz, 2003; Scholz et al., 2003; Scholz and Schöner, 1999; Yang et al., 2007; Yen and Chang, 2010). Less consideration has been given to gait and the relative motor redundancy it contains. Among the many studies that looked into gait and centre of mass stabilisation, only one used the UCM approach (Black et al., 2007) and one used covariation analysis, a method comparable to the UCM (Verrel et al., 2010). Both studies however limited their analysis to walking at times of heel strikes rather than the entire time history of the gait cycle. The ability to analyse the whole cycle rather than a single instant would have greater clinical impact in rehabilitation of impaired gait. Of the few studies (Krishnan et al., 2013; Robert et al., 2009; Rosenblatt et al., 2014) that analysed the temporal evolution of the UCM approach throughout the gait cycle, none have considered the COM trajectory as a performance variable. The exploratory study reported here, therefore, investigated the potential usefulness of UCM analysis of postural control during the stance phase of walking. It is known that stabilisation of the COM is key to walking ability so that is the component of postural control identified as the key focus.

We hypothesised that different combinations of lower limb joint angles (kinematic synergy) can be used to control the COM movement while walking. The specific aims of this study were to (a) determine the feasibility of undertaking an UCM analysis of control of the COM during the stance phase of gait, (b) to find if such analysis could provide more knowledge of COM control than 'standard' biomechanical analysis techniques, (c) explore whether the UCM analysis could identify differences between a stroke survivor with walking difficulty and adults without a brain lesion and the stroke survivor walking with and without a custom-made ankle-foot orthosis (AFO).

2. Methods

2.1 Participants

Six adults (3 female, 3 male; height: 168.9 (± 10.5) cm, mass: 68.2 (± 9.9) kg, age: 29.8 (± 6.7) years) with no known neurological pathology participated in the study. In addition, one 81 years old male (80 kg, 180 cm) was recruited 2 months after experiencing a stroke. He presented a left side hemiplegia of the upper and lower body.

He was prescribed a 5mm polypropylene AFO with carbon fibre reinforcement at the malleoli level. The AFO and shoes combination were tuned at 10° of forward inclination. All participants provided written consent for the study which was approved by the local ethics committee (West of Scotland REC3).

2.2 Equipment and Experimental Procedure

A twelve-camera motion capture system (Vicon, Oxford Metrics Ltd., UK) was used to collect experimental data at 100 Hz while participants walked at comfortable speed on a flat surface of 6 m in length. The gait analysis protocol developed within the Bioengineering Department at University of Strathclyde was followed for data collection and processing (Papi et al., 2011).

Ten walking trials were recorded with able-bodied subjects and the data derived from their left leg were used in the subsequent analysis. The stroke participant was assessed three times at three monthly time points. Six trials were collected during walking with and without AFO at each visit. Data from the hemiplegic leg were considered.

2.3 Data Processing

Initial data processing was performed using Nexus software (Oxford Metrics Ltd., UK). Hip, knee and ankle sagittal angles, and the 3-D coordinates of anatomical landmarks were output. Data were time normalised to 100% of stance phase.

2.4 UCM Formulation

The UCM method was applied to characterise the control of the COM during stance phase. The performance variable was COM movement and the elemental variables were the lower limb joint rotations. As a preliminary development of the UCM method for gait, it was decided to conduct the analysis for stance phase in the sagittal plane and to approximate the COM as a fixed point in the pelvis. This point was defined by the intersection of the diagonals connecting the anterior and posterior superior iliac spines. To estimate how the variability of joint angles influences the position of the COM in a global sagittal plane with x antero/posterior and y vertical axis, a geometric model (Fig. 1) that links hip, knee and ankle rotations to the COM position throughout the gait cycle was defined. Since the position of the COM depends also on the position of the foot on the ground and in particular on the angle between the sole of the foot and the ground, this angle was also considered an elemental variable:

$$(x_{COM}, y_{COM}) = f(\theta_G, \theta_A, \theta_K, \theta_H) \quad (1)$$

Where: θ_G is the angle between the sole of the foot and the ground, θ_A, θ_K θ_H are the ankle, knee and hip sagittal angles respectively. θ_G is the angle (Fig. 2) between the vectors $\vec{a}$, characterising the sole of the foot, and $\vec{b}$ representing the ground:

$$\theta_G = \arcsin(\vec{a} \times \vec{b}\,) \quad (2)$$

The first modelling requirement was to consider the position of the foot on the ground during walking to define the ankle joint centre. Three main cases were identified (Fig. 2):

1- the heel is in contact with the ground: $\theta_G > 0$,

2- foot flat: $\theta_G = 0$,

3- the fore part of the foot is on the ground: $\theta_G < 0$.

The sagittal position of the ankle joint centre was defined as follow, for case 1 and 2:

$$x_A = x_{Calcaneus} + CA\cos(\alpha + \theta_G) \quad (3)$$

$$y_A = y_{Calcaneus} + CA\sin(\alpha + \theta_G) \quad (4)$$

Where: α is the angle at the rear of the foot (Fig. 2) and CA is the length of the segment joining the ankle joint centre (AJC) and the calcaneus.

For case 3:

$$x_A = x_{Midpoint\ 1st\ and 5th} - MA\cos(\theta_G - \beta) \quad (5)$$

$$y_A = y_{Midpoint\ 1st\ and 5th} - MA\sin(\theta_G - \beta) \quad (6)$$

Where: β is the angle at the fore part of the foot (Fig 2) and MA is the length of the segment joining the midpoint between the 1st and 5th metatarsal head and the AJC.

Through a trigonometric analysis of the leg segment, the COM position in the sagittal plane can be expressed as:

$$x_{COM} = x_A + AK\cos(\theta_G + \theta_A + \pi/2) + KH\cos(\theta_G + \theta_A + \theta_K + \pi/2) + HCM\cos(\theta_G + \theta_A + \theta_K - \theta_H + \pi/2) \quad (7)$$

$$y_{COM} = y_A + AK\sin(\theta_G + \theta_A + \pi/2) + KH\sin(\theta_G + \theta_A + \theta_K + \pi/2) + HCM\sin(\theta_G + \theta_A + \theta_K - \theta_H + \pi/2) \quad (8)$$

Where: AK is the shank segment length, KH is the thigh segment length, HCM is the hip centre to COM segment length.

The next step required for this approach was the linearization of the UCM (Latash et al., 2007). This was necessary because the concept of variance is a linear concept while, the UCM, and in particular the geometric model defined, are not linear. The linearization implies the definition of the Jacobian matrix, $J(\theta)$, and the computation of its null space, N(J). The Jacobian matrix is a matrix of all first-order partial derivatives of the COM coordinates with respect to the elemental variables. Changes in joint angles and changes of the COM trajectory are linked through this matrix.

The null space (Eq. 9) of the Jacobian matrix, spanned by the basis vectors ε_{n-d}, is the linear subspace of all joint angles combinations that leave the COM coordinates unaffected. The dimension of this subspace is *(n – d)* where *n* is the number of elemental variables and *d* is the number of dimensions of the performance variable. The null space in the current case had a dimensionality of 2.

$$0 = \boldsymbol{J}(\boldsymbol{\theta}) \cdot \varepsilon_{n-d} = \begin{bmatrix} \frac{\partial x_{COM}}{\partial \theta_G} & \frac{\partial x_{COM}}{\partial \theta_A} & \frac{\partial x_{COM}}{\partial \theta_K} & \frac{\partial x_{COM}}{\partial \theta_H} \\ \frac{\partial y_{COM}}{\partial \theta_G} & \frac{\partial y_{COM}}{\partial \theta_A} & \frac{\partial y_{COM}}{\partial \theta_K} & \frac{\partial y_{COM}}{\partial \theta_H} \end{bmatrix} \cdot \varepsilon_{n-d} \longrightarrow \boldsymbol{N}(\boldsymbol{J}) = \begin{bmatrix} \varepsilon_{11} & \varepsilon_{12} \\ \varepsilon_{21} & \varepsilon_{22} \\ \varepsilon_{31} & \varepsilon_{32} \\ \varepsilon_{41} & \varepsilon_{42} \end{bmatrix} \quad (9)$$

The linearization was performed around a reference configuration, defined as the mean joint configuration across trials for each instant analysed (Latash et al., 2007). This is assumed to represent the set of angles that leads to the desired COM position. The Jacobian matrix, $J(\bar{\theta})$, was calculated with respect to this configuration. The computation of the Jacobian matrix and then of its null space was performed for each time point of stance phase of each trial and hence they continuously varied.

In addition, the deviation, for each trial, from the mean joint configuration ($\bar{\theta}_G, \bar{\theta}_A, \bar{\theta}_K, \bar{\theta}_H$) was calculated at each instant:

$$\boldsymbol{DV} = \begin{bmatrix} \theta_G - \bar{\theta}_G \\ \theta_A - \bar{\theta}_A \\ \theta_K - \bar{\theta}_K \\ \theta_H - \bar{\theta}_H \end{bmatrix} \quad (10)$$

The obtained deviation vector (DV) was decomposed into a component that is within ($\theta_{\parallel}$) and perpendicular ($\theta_{\perp}$) to the null space:

$$\theta_{\parallel} = \sum_{i=1}^{n-d} \left(\boldsymbol{N(J)}_{\boldsymbol{i}}^{\boldsymbol{T}} \cdot \boldsymbol{DV}\right) \boldsymbol{N(J)}_i \qquad (11)$$

$$\theta_{\perp} = \boldsymbol{DV} - \theta_{\parallel} \qquad (12)$$

The scalar values obtained represent to what extent the trial joint configuration is consistent with the reference joint configuration.

The variances of these projections ($\theta_{\parallel}$, $\theta_{\perp}$) were then calculated. Since they have different dimensions, the variances were normalised per degree of freedom of each subspace. The variance across trials within the linearized UCM ($\sigma_{\parallel}^2$) and perpendicular to it ($\sigma_{\perp}^2$) are:

$$\sigma_{\parallel}^2 = \frac{\sum_{i=1}^{N} \theta_{\parallel N}^2}{(n-d)N} \qquad (13)$$

$$\sigma_{\perp}^2 = \frac{\sum_{i=1}^{N} \theta_{\perp N}^2}{dN} \qquad (14)$$

Where: $\theta_{\parallel}^2$ and $\theta_{\perp}^2$ are the squared length of the deviation vector component lying within and perpendicular the UCM respectively, *N* is the number of trials, n are the elemental variables and *d* is the dimensionality of the performance variable.

The variances within and perpendicular the UCM were compared to verify the initial hypothesis about the control of the COM trajectory during walking.

A ratio defined as in (Eq. 15) was used to summarise the results:

$$Ratio = \left(\frac{2\sigma_{\parallel}^2}{\sigma_{\parallel}^2 + \sigma_{\perp}^2}\right) - 1 \qquad (15)$$

2.5 Data Analysis

Matlab (The MathWorks Inc., Massachusetts, US) was used for data processing. Data analysis was conducted by first analysing the variability of joint angles and COM trajectories and then, by evaluating the structure of variances within the UCM framework.

For the analysis, stance phase was divided into three phases (i) contact phase (20%, initial contact and loading response), (ii) mid stance (30%), (iii) propulsive phase (50%, Terminal stance and pre swing) (Perry et al.,) to allow a more detailed analysis of the evolution of variances components during stance. To test the hypothesis that a kinematic synergy of the joint angles stabilise the COM trajectory and to evaluate if the strength of the synergy change within stance phase, a repeated measures ANOVA was performed with factors of variance components ($\sigma_{\parallel}^2$ and $\sigma_{\perp}^2$) and phases (contact phase, mid stance, propulsive phase).

One-tailed t-test was applied to verify if the ratio was significantly greater than 0, implying $\sigma_{\parallel}^2 > \sigma_{\perp}^2$, and hence accepting the hypothesis about the control of the COM movement during gait. Two-sample t-test was used for comparisons between stroke patient and able-bodied subjects parameters. Significance was set at 0.05.

Prior to statistical analysis, variances components and ratios were averaged across each phase of the stance phase.

3. Results

Overall sagittal joint kinematic mean pattern and variability (standard deviations bars) throughout stance phase are shown in Figure 3 for each able-bodied subject and in Figure 4 for the hemiplegic leg of the stroke participant during 3 assessments for walking with AFO and without AFO. Higher variability in joint kinematic is observed for the stroke participant and in particular during baseline test.

COM displacements in the x and y directions are illustrated in Figure 5 and 6 for able-bodied participants and stroke patient respectively. The latter showed higher variability in COM trajectories.

The variability in joint kinematics was related to the variability of the COM position through the UCM analysis. Time series of the variance within and perpendicular to the linearized UCM are presented in the graphs on the right side and ratios on the left side of Figure 7 and 8 for able-bodied participants and stroke patient respectively. Ratio values above or equal to 0 indicate that the hypothesis about the control of the COM can be accepted. Ratios for able-bodied participants and the stroke patient at the three assessments were all statistically different from 0 (p<0.000) and no statistical difference existed between stroke and able-bodied participant ratios (p=0.216).

The results of the ANOVA indicated a significant main effect of variance component ($\sigma_{\parallel}^2 > \sigma_{\perp}^2$) supporting the hypothesis that a kinematic synergy existed to stabilise the COM ($F_{1, 11}$=18.5; p=0.001) but it showed that the synergy does not change within different periods of the stance phase ($F_{2, 22}$=1.1; p=0.330). The stroke patient despite showing variability in joint angles higher without the AFO (Fig. 4) and in comparison to able-bodied

adults, maintained a stabilised COM position during walking, with and without the AFO. $\sigma_{\parallel}^2 >$ was statistically different between stroke and able-bodied participants, although $\sigma_{\perp}^2$ did not reach significance.

The control of COM when the patient used the AFO was reduced over time (from first to last visit); on the contrary, the control imposed without the AFO was always high. Confidence and stability acquired when walking with the AFO allowed the patient to be less vigilant with regards to COM displacement.

For the patient a progression towards the ratio waveform seen in able-bodied subjects can be observed particularly at the 6-months follow-up, although differences were present due to altered gait events timing in hemiplegic walking (i.e., Prolonged terminal stance in which the hemiplegic leg waited for the contralateral foot to completely strike the ground before going into swing phase).

4. Discussion

4.1 Interpretation of UCM method findings

The UCM hypothesis proposes that variability in movement patterns is essential for the performance of everyday functional tasks in a variety of environmental contexts. In other words, what was termed DOF problem might actually be a DOF advantage especially in the presence of a stroke lesion in the brain. Availability of different movement patterns to achieve the same functional goal enables achievement of that goal albeit with what is consider sub-optimal movement. Utilising the UCM approach, in this study, the variability obtained in the COM displacements in the sagittal plane during stance phase depending on sagittal kinematics was classified accordingly to the subject's ability to control the position of the COM. Lower limb sagittal kinematics showed variability through stance phase that was more evident in the stroke participant and particularly for walking without an AFO. All participants had, overall, a variance within the UCM greater than its complement in all phases of stance phase and hence maintained the COM position despite the stroke patient presenting with a higher variability of joint kinematics. This indicated that the patient adopted a way of walking which utilised more variation in joint configurations (Fig. 4) without altering the COM position.

A tendency to increase the variance within the linearized UCM was observed when the patient walked without the AFO, indicating a greater focus was placed on the mainteinance of the COM position than when walking with an AFO. The AFO appeared to give the patient more confidence such that he was able to lower the control imposed on the COM position and approach a more normal gait.

The method highlights how adults with and without brain lesion used differently the available DOFs at the joint level but all achieving a stable COM movement during stance of the gait and how this can change by the use of AFO and with time.

The aim of applying the UCM approach was to introduce a method, that has been applied to more stationary tasks, to gait in order to improve our understanding of how able-bodied and impaired populations control walking and thus providing additional explanation to standard kinematics and trajectories curves. This approach provides explanation of the variability in joint kinematics and how these are functionally employed with the purpose of achieving a successful task performance by controlling a particular parameter, in this case the trajectory of the COM. In rehabilitation practice it could guide clinicians as to how intervene and verify if selected interventions hinder or improve the achievement of locomotion by identifying which movement patterns should be encouraged to achieve a succesfull task performance.

4.2 Methodological Considerations

The UCM approach was formulated to test the hypothesis of the control of COM position during walking. The advantages of the method proposed over studies already conducted and looking into COM trajectory are the ability to perform the analysis through a period of time rather than only instances of the motor task performed and its applicability to gait whereas most of the previous studies concentrated on upper body performances with only few in lower limbs tasks and gait (Auyang et al., 2009; Black et al., 2007; Domkin et al., 2002; Hsu et al., 2007; Krishnan et al., 2013; Reisman and Scholz, 2003; Robert et al., 2009; Rosenblatt et al., 2014; Scholz et al., 2003; Scholz and Schöner, 1999; Yang et al., 2007; Yen and Chang, 2010) .

The method introduced showed to be feasible as reasonable and interpretable results were obtained although not without limitations. The geometric model used assumed the COM as a fixed point in the pelvis without taking into account the contributions of the arm and trunk movement, the analysis was confined to the sagittal plane and to the duration of stance phase. These choices were made to simplify the formulations of the UCM approach in the first instance. Calculation of the true COM as a sum of each body segment centre of mass weighted with respect to segments' mass could be included as further development of this approach as well as the COM 3D coordinates. The analysis in swing phase is more complex as no fixed support is available for the swinging leg. A further development could be increasing the complexity of the model adopted by accounting for DOFs that were not considered in the current study. For example considering both legs will give a better description of the

3D COM movement and will overcome the lack of a fixed base in swing as the analysis will switch to the contralateral leg.

5. Conclusion

The feasibility of the application of the UCM method to gait data time series was shown in both adults with and without neuropathology. Further research on a larger population group is required to strengthen the clinical meaning of the findings. However, this approach shows great promise with respect to understanding postural control mechanism during walking and the relation to rehab approaches.

The UCM analysis allows a classification of the variability of a selected control variable with respect to the elemental variables and it can be seen as a tool to appraise the effect of rehabilitative techniques and rehabilitation over time. Its utility is to provide an explanation of how the CNS acts to cope with the different DOFs available for the walking task and how it compensates if impairments at the elemental variable level are present. It is another form of analysis to be added to the results of 3-D gait analysis in the process of evaluating stroke gait kinematics for which a retrospective analysis is conducted through the UCM.

Conflict of interest

We declare that we have no conflict of interest

Acknowledgement

Nonthing to declare

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Figures Captions:

Fig. 1. Leg and foot stick model.

Fig. 2. Positions of the foot on the ground at the three identified key points depending on θ_G value.

Fig.3. Mean and standard deviations shade areas of sagittal hip, knee and ankle kinematics of six able-bodied subject over 10 walking trials during stance phase.

Fig. 4. Mean and standard deviations shade areas over 6 gait cycles of sagittal hip, knee and ankle joint kinematics with (dashed lines) and without AFO (solid lines) for the stroke patient during the three assessments.

Fig. 5. Mean COM displacement during stance in antero/posterior (x) and vertical (y) directions for the six able-bodied subjects. Standard deviation shade areas are shown.

Fig. 6. Mean COM displacement during stance in x and y directions for the stroke patient during walking with (dashed lines) and without AFO (solid lines). Standard deviation shade areas are shown.

Fig. 7. Variance components within (Vucm, solid lines) and perpendicular (Vort, dashed lines) to the linearized UCM for able-bodied subjects (left). Mean ratio (± standard deviation) across six able-bodied subjects (ratio).

Fig. 8. Variance components within (Vucm, solid lines) and perpendicular (Vort, dashed lines) to the linearized UCM and mean ratio for the stroke patient at three assessments with (solid lines) and without AFO (dashed lines).