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# Motor learning without the sixth sense

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## INTRODUCTION:

One of the most intriguing issues in behavioral neuroscience is the extent to which the brain selects sensory signals for performing motor tasks. Recently, Sober and Sabes (2005)[1] suggested that sensory signals are processed differently according to the nature of the task and the required neural processing. Their computational model indicated that visual signals are mainly used to specify a desired movement, in Cartesian coordinates, while proprioceptive signals from the muscles, joints and skin are critical to determine intrinsic, physiologically-relevant motor commands. Although proprioception is not listed among the five senses, its role is critical for movement control, as demonstrated by empirical observations on healthy humans and most strikingly on proprioceptively-deafferented patients[2]. These rare patients exhibit severe movement impairments and pinpoint how important the (sixth) proprioceptive sense is for adapting motor commands to the physical constraints acting on the limbs. In contrast, the role of the visual sense is still considered negligible for such adaptation as blindfolded subjects, and even congenitally-blind subjects, can adapt their pointing hand movements to novel force fields[3]. The present study[4] was designed to further examine the hypothesis that vision can be useful to model limb dynamics when adapting to a novel force field.

## METHOD:

We had the unique opportunity to study the motor behaviour of a proprioceptively-deafferented patient on a rotating platform. Patient G.L., who suffers from a complete loss of proprioceptive afferents from nose to toe, provides an opening to study the role of visual information in motor control without the confounds related to the presence of proprioceptive information. The rotating platform enabled us to create a non-terrestrial, artificial gravity environment and to generate continuous forces which affect movement execution. We thus asked the patient to point toward visual targets before, during and after the platform rotation to assess her ability to adapt pointing movements to the novel force environment. Six age-matched, control participants were also tested.

We used a Codamotion optical motion tracking system to record, at 500 Hz, the position of the fingertip and thus measure pointing accuracy. A single cx1 system with Mini Hub was embarked on the platform which could rotate up to 120°/s. We controlled the motion tracker, the motorized platform and the presentation of the visual targets from an adjacent room by using a customized software (Docometre; www.ism.univmed.fr/~buloup/) governing a real-time acquisition system ADwin-Pro (Jäger, Lorsch, Germany; www.adwin.de) which was also embarked on the platform.

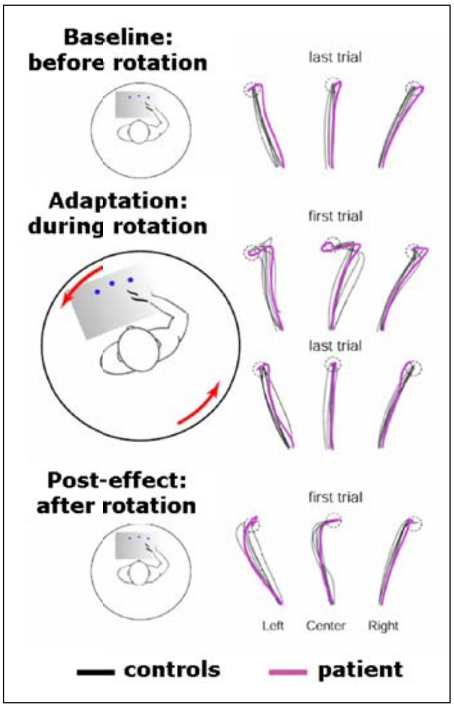


***Figure 1.*** *Top view of the hand paths of the patient and all control participants before, during and after platform rotation.*

## RESULTS:

The middle panel of Figure 1 shows that the deafferented patient was able to adapt to the novel force environment as efficiently as control participants (see[4] for statistical analyses). Moreover, we observed similar post-effects after the rotation ended (lower panel of Figure 1), suggesting that the patient could develop and store in memory, as well as controls, an internal model of the novel force environment and adapt her pointing movements.

## CONCLUSION:

Our study thus provides clear evidence that the deafferented patient could learn the novel dynamic environment by using visual feedback. This contrasts with the classical view that proprioception is required to adapt to the musculoskeletal and environmental physical constraints. Our findings provide direct evidence that vision can compensate for the permanent loss of proprioception when learning new dynamic conditions, highlighting the brain's capacity to process different sensory inputs to obtain the same functional result. Our results thus question the specificity of the senses for motor learning and support the idea of sensory substitution for sensorimotor rehabilitation[5].

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# A portable system for collecting anatomical joint angles during stair ascent: a comparison with an optical tracking device

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## BACKGROUND:

In terms of self-rated health, the most important activities of daily living are those involving mobility. Self-reported difficulty in stair climbing has shown to be useful in assessing and defining functional status of older adults. Obtaining accurate data about mobility is therefore of great clinical relevance and could lead to further improvements in various rehabilitation treatments. In general, kinematics and biomechanical aspects of stair climbing are studied using laboratory staircases combined with an optical motion analysis system [1-2]. Although this kind of research yields valuable information, the results only remain valid in conditions where no anticipation or reaction to a real-world environment is required. In addition, it is almost impossible to use any form of optical tracking on stairwells, as the vertical shaft which contains the staircase limits the placement of cameras. Collecting data during stair climbing in a more real-life, complex environment requires a portable and lightweight measuring device. Zhou et al (2006) showed that inertial measurement units (IMUs) consisting of gyroscopes, accelerometers and magnetometers used to measure upper limb motion can accurately estimate arm position [3]. Although, a miniature gyroscope attached to the shank is able to detect different cycles during stair ascent [4] and position data of the foot can be gathered with the combination of a gyroscope and two accelerometers [5], a portable system that can collect anatomical joint angles during stair climbing has not yet been reported. The purpose of this study is to compare the anatomical joint angles determined by IMUs during stair ascent, to those joint angles acquired with an optical tracking device.

## METHODS:

Fourteen healthy subjects with a mean age of 27 years (range 20 to 37) participated in this study after consent was given. Each subject was asked to ascend a staircase consisting of four steps during twelve separate trials. Subjects were instructed to climb the stairs in the way they felt most comfortable.

Six IMUs (MTx, Xsens Technologies B. V., Enschede, Netherlands) were placed on the dorsal side of both forefeet , halfway up the medial surface of the tibias and two thirds up the tensor fascia latae of each leg using double-sided adhesive tape with additional elastic straps to hold them in place (Figure 1).

During static stance, the X-axis of each IMU coordinate system was physically placed to be in the sagittal plane after an analytically alignment of the axes by software (MT Software V2.8.1, Xsens Technologies B. V., Enschede, Netherlands). The software program placed the Z-axis of each IMU in line with gravity (vertical plane) with the new X-axis of the sensor perpendicular to the Z-axis and along the line of the original X-axis.

Active Codamotion (Codamotion, Charnwood Dynamics, Leicestershire, UK) markers were placed (Figure 1) on the toe (5th metatarsal head), ankle (lateral malleolus), knee (fibula head and lateral femoral condyle), hip (trochanter major) and on the side of stairs.

*__Figure 1.__ Sensor set up used during study. Optical tracking markers and Inertia Measurement Units (IMUs) as attached to each subject.*

The Bilateral Segmental Gait Analysis system configuration was used for data acquisition by the Codamotion and Motion Tracker software. Data for both the Codamotion and the IMUs was acquired at 100 Hz and an electronic pulse was used to synchronize the two measurement devices. All further data analysis was done using Matlab (MathWorks, Inc., Natick, MA, USA).

## DATA ANALYSIS:

The lower extremity could be approximated as a multi-link chain, with each body part as a rigid segment represented by one IMU [6]. Only movements around the transverse axis (resulting in flexion-extension kinematics) were studied.

The rotation matrix (RDCM), which was acquired from each IMU, was used to determine the Euler angle (θ) that represented rotation around the transverse axis. The angle (θ) for each of the six IMUs combined with segment lengths of the foot, shank and thigh were used in a six-link sagittal model (Figure 2).

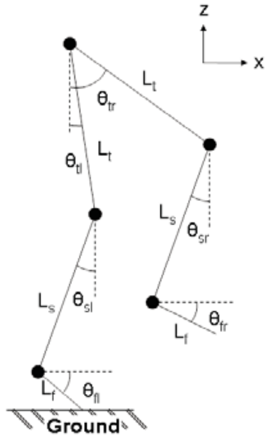


*__Figure 2.__ Six-link sagittal model. (Lf) length of the foot; (Ls) length of the shank; (Lt) length of the thigh; (θfl) angle of the left foot; (θsl) angle of the left shank; (θtl) angle of the left thigh; (θfl) angle of the right foot; (θsl) angle of the right shank; (θtl) angle of the right thigh.*

The segment lengths were calculated from anthropometric data [6]. The knee angle (α) was determined with the IMUs, by subtracting the angle around the transverse axis of the shank from that of the thigh. The flexion-extension angle of the ankle was found by subtracting the lower leg angle from the foot angle in the sagittal plane, while the thigh angle was represented by the upper leg angle with respect to the vertical axis [7].

For the optical tracking device the knee angle was defined as the angle between a spatial vector joining the lateral malleolus to the fibula head and a spatial vector joining the lateral femoral epicondyle to the greater trochanter [8]. This calculation method described by Kiss, Kocsis and Knoll determines a knee angle (α) which only depends on the relative position of the shank to the thigh. The same principle can be applied for the determination of the ankle angle (β), by using the spatial coordinates of the most lateral aspect of the calcaneus (x2, y2, z2) and the 5th metatarsal head (x1, y1, z1). The thigh angle (γ) was defined as the angle between a spatial vector joining the epicondylus fermoris lateralis and the trochanter major and a vertical spatial vector.

All angles were normalized in time per trial and subject by calculating the mean angle per percentage of time.

## STATISTICAL ANALYSIS:

Data was normally distributed as observed in the probability plots and histograms. All anatomical joint angles on the right leg, obtained with the two measurement devices, were evaluated by calculating a two-tailed Pearson product-moment correlation coefficient (r) and by calculating the root mean square error (RMSE) between the two signals [9]. A paired t-test was used to compare the maximum range of motion obtained by IMUs with those obtained by the optical tracking device per subject (n=14) and to determine if the slopes of the linear regressions differed from one.

## RESULTS:

During a hundred trials subjects took their first step with the right foot, while they started with the left in 68 trials (Figure 3).

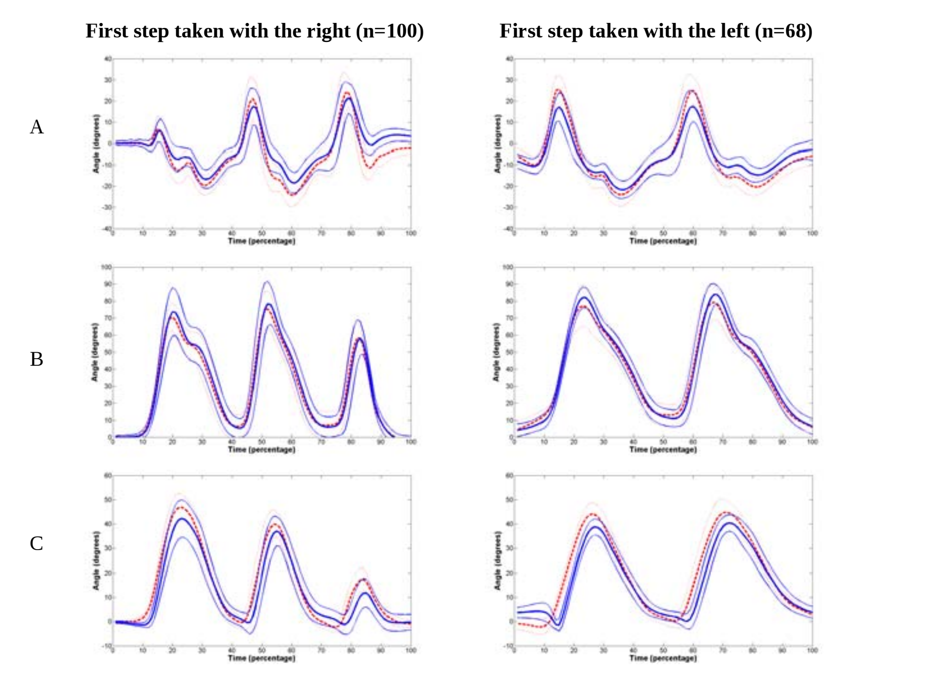


***Figure 3.*** *Mean angles and standard deviation of the right leg in the sagittal plane. Thick red dotted lines are the mean angles obtained by IMUs and the thick blue solid lines are those obtained by the optical tracking device. Thin lines represent the standard deviations. A: Ankle, B: Knee, C: Thigh.*

High correlation coefficients and low RMSE were found (Table 1). A significant difference was found between the maximum range of motion of the ankle and thigh ($p<0.01$) obtained with the IMUs compared to those acquired with the optical tracking device. No difference was found for the maximum range of motion at the knee joint ($p=0.47$).

**Table 1** - Pearson correlations, Root Mean Square Errors and maximum Range of Motions

| | *Pearson correlation coefficient (p<0.01)* Mean (SD) | *Root Mean Square Error in degrees* Mean (SD) | *Maximum Range of Motion in degrees* Mean (SD) IMUs | Optical device |
|---|---|---|---|---|
| **Ankle angles** | 0.93(± 0.05) | 4 (± 2) | 63 (± 8) * | 54 (±8) |
| **Knee angles** | 0.98(± 0.05) | 4 (± 3) | 91 (± 8) | 92 (±6) |
| **Thigh angles** | 0.96(± 0.06) | 5 (± 3) | 56 (± 5) * | 49 (±4) |

Mean correlations and Root Mean Square Errors between joint angles in the sagittal plane, acquired by the IMUs and the optical tracking device over all the 14 subjects and 12 trials (n=168). Maximum Range of Motion obtained with both measurement devices per subject (n=14). SD: standard deviation. Asterisks indicate difference ($p<0.01$) between maximum Range of Motion with respect to the optical tracking device.

## DISCUSSION:

The aim of the study was to investigate if the anatomical joint angles determined by IMUs sufficiently approximate the anatomical joint angles that were gathered with an optical tracking device. Strong correlations and mean RMSE of 4 to 5 degrees were found for all angles, comparable to those obtained using a similar system to track upper limb motion [10].

In the unpublished pilot study to this paper, which compared the equipment component of both systems, Pearson's correlation coefficients of 0.999 ($p<0.001$) between the IMUs and optical tracking device were found, with a RMSE of 1°. As the RMSE during stair climbing (4-5°) was greater than the RMSE found in the pilot study (1°), a small misalignment between the two coordinate systems could have been present.

Both systems suffer from motion artefacts. A translational displacement between bony landmark and marker is likely to occur during stair climbing, causing errors in estimating position during movement [11-12], which in turn leads to inaccuracies in determining angles. The IMUs measure orientation rather than position and are consequently less prone to errors caused by translational displacement of the sensors. Errors related to movement can however still occur in the IMUs, because of rotational displacement of the sensor relative to the body segment.

Maximum range of motion of the knee angle was similar between the two measurement devices, but did differ in the thigh and ankle angles. Some further clues about the differences found between the two systems are provided by inspection of individual traces which deviate strongly from the rest (Figure 3). Data inspection showed that these deviations occurred in the optical tracking marker position, presumably due to movement of the marker with respect to the bony landmark.

In general, IMUs provide a good alternative for measuring joint angles of the lower extremity during stair ascent when compared to positional markers. In addition, they provide the opportunity to perform accurate measurements in complex real-life environments using a non constraining measurement device. Furthermore, the IMUs were easy to set up, giving rise to the opportunity for clinicians and researchers to measure stair climbing out of the laboratory setting.

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# The influence of work rate and cadence on movement coordination in cycling

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## INTRODUCTION:

In a kinematic chain the motion of one segment subsequently influences the motion of an adjacent segment, and therefore the study of isolated joints does not effectively capture the complexity of the coordinated motion of components of the body (Bartlett et al., 2007). The consideration of the coupling relationship between segments may therefore be crucial in the analysis of human movement. There is conflict within the cycling literature regarding the most economical cadence, definedin this study as that which is associated with the lowest metabolic cost at a given work rate. This is due in part to its work rate-dependent nature (Ansley & Cangley, 2009). Li (2004) found as cadence increases there is an added influence of the inertial properties of the limbs, which consequently affects neuromuscular coordination. Changes in the coordination patterns utilised by cyclists as a result of changes to the work rate and /or cadence may therefore have an effect on their economy. A key component in the analysis of movement coordination is the role of variability within the system under investigation (Wilson et al., 2008). Movement variability is important in skills where the adaptability of complex motor patterns is necessary within dynamic performance environments (Button et al., 2006), enabling athletes to adjust to both intrinsic and extrinsic factors (Bradshaw & Aisbett, 2006). However, in skills where tight task constraints are imposed, such as in cycling, there is likely to be a reduced requirement for flexibility and any variability present in the system may therefore be indicative of an inconsistent performance. In support of this Chapman et al. (2009) concluded that elite cyclists had greater consistency of inter-joint coordination compared with novice cyclists. The aim of this study was therefore to investigate the affect of the work rate and cadence on the coordination exhibited by trained male cyclists and the subsequent implications for training and competition in terms of adopting the most economical strategy.

## METHODS:

Six trained male cyclists were recruited for the study. All subjects gave written informed consent and were free from injury at the time of the study. Using a two-scanner Cartesian Optoelectronic Dynamic Anthropometer (CODA) motion analysis system three dimensional kinematic data were collected at a sampling rate of 100 Hz. Exercise was performed on a Monark braked cycloergometer. Twenty-three active markers of 2-mm diameter were attached to the right lower limb and the pelvis. The markers were located on the following anatomical landmarks: 5th metatarsal head, 1st metatarsal head, lateral malleolus, medial malleolus, heel, medial and lateral knee epicondyles, greater trochanters, anterior superior iliac spines, iliac crests and posterior superior iliac spine. The remaining markers were attached to polystyrene plates which were placed on the distal thigh and shank. Each plate contained a cluster of 4 markers. An additional marker was placed on the pedal axis in order to identify individual revolutions. Subjects undertook 9 pedalling bouts in a randomized order at various work rates and cadences (120, 210, 300 W at 60, 90, 120 rpm). Subjects were instructed to reach the required cadence (visual feedback provided by digital RPM-meter) and maintain this for at least 10 seconds before data recording commenced. Data were recorded for a minimum of 20 s. A minimum of a one minute of recovery was given between trials. For each trial a total of 10 consecutive revolutions within ± 2 rpm of the required cadence were selected for subsequent analysis. A complete revolution was the time from top dead centre (TDC) to subsequent TDC. TDC was defined when the pedal marker reached its maximal value in the z-axis. Visual 3D motion analysis software (C-motion) was used to calculate 3D joint angles according to a method outlined by Grood and Suntay (1983). Prior to this raw coordinate data was smoothed using a fourth order Butterworth digital filter with a cut-off frequency of 8 Hz. The cut-off frequency was selected using Winter's (1990) residual analysis technique. Only sagittal plane data were used for further analysis. The time series of each joint angular position and velocity was assessed on a revolution-by-revolution basis and interpolated to 100 data points using a cubic spline technique. The intra-limb joint coupling motions were assessed for each revolution using a continuous relative phase (CRP) analysis, which was calculated using the angular position and velocity profiles of the relationship between the joint actions (Dierks and Davis, 2007). CRP was assessed for 2 intralimb couplings: ankle plantarflexion/dorsiflexion - knee flexion/extension (KA) and knee flexion/extension – hip flexion/extension (HK). The joint angle and angular velocity data were normalised to the maximum and minimum of the athlete-specific data set according to the procedure presented by Hamill et al. (1999). The CRP time histories for the sagittal plane KA and HK joint couplings were determined by quantifying the difference between the phase angle of the distal and proximal joint at each time interval. CRP describes the relationship between two oscillators in the phase-plane domain. A CRP of $0^{\circ}$ indicates in-phase coupling, meaning the phase angles for the two motions are identical, and a potentially stable coupling pattern exists as they are behaving similarly. As the CRP increases from $0^{\circ}$ in either a positive or negative direction, the two motions become more out-of-phase and are behaving in a less similar fashion. Individual averaged time histories for the CRP and the associated variation of CRP (CRPv) were determined across all revolutions for each trial using the mean CRP and associated standard deviation (SD) respectively at each time point. Time histories for the group averaged CRP and CRPv were determined as the average across each time point of the individual-specific CRP and within athlete CRP averaged profiles, respectively. This was repeated for each condition. For each coupling, the effects of cadence and work rate (and the subsequent interaction effects) on CRP and CRPv were determined using a 2-way repeated measures ANOVA. Where significant interaction effects were identified, post hoc analyses were employed to examine where the significant differences existed. In addition, differences in CRP and CRPv between the propulsive and recovery phases of the revolution were examined. Significant differences were accepted at $p < 0.05$.

## RESULTS:

No significant differences in CRP or CRPv were found between work rate conditions for either KA or HK. Significant differences in CRP were found between the propulsive and recovery phases for both couplings with a more in phase motion being displayed during the propulsive phase (propulsive vs recovery; KA, $27.4^{\circ} \pm 8.9$ vs $48.5^{\circ} \pm 20.5$, $p = 0.000$; HK, $22.5^{\circ} \pm 6.7$ vs $32.5^{\circ} \pm 6.8$, $p = 0.000$). Significant differences in CRP were also found between the cadences for the HK coupling during the recovery phase with the 60 RPM trial displaying more out of phase motion than either the 90 RPM or 120 RPM trials ($36.4^{\circ} \pm 3.5$ for 60 RPM vs $33.3^{\circ} \pm 3.4$ for 90 RPM, $p = 0.030$ and $27.9^{\circ} \pm 13.6$ for 120 RPM, $p = 0.026$). Differences in CRP for the KA coupling were found during the propulsive phase only with the 120 RPM trials displaying significantly more in phase motion than either the 60 RPM or the 90 RPM trials ($19.2^{\circ} \pm 12.3$ for 120 RPM vs $30.0^{\circ} \pm 7.1$ for 60 RPM, $p = 0.011$ and $33.1^{\circ} \pm 7.4$ for 90 RPM, $p = 0.024$). There were no differences in CRPv across the cadence conditions for the HK coupling however in the KA coupling a significantly higher CRPv was displayed during the recovery phase in the 60 RPM trials compared to either the 90 RPM or 120 RPM trials ($16.6^{\circ} \pm 7.6$ for 60 RPM vs $11.6^{\circ} \pm 6.5$ for 90 RPM, $p = 0.005$ and $8.9^{\circ} \pm 4.1$ for 120 RPM, $p = 0.003$).

## DISCUSSION:

The intra-limb coupling motion of trained male cyclists was quantified for the propulsive and recovery phases of cycle revolutions at three different work rates (120, 210 & 300 W) and three different cadences (60, 90 & 120 RPM). The more out of phase motion of both the KA and HK couplings during the recovery phase suggests a less stable motion than in the propulsive phase as out of phase motion has previously been considered to reflect a less stable coordinative state (Scholz, 1990). When considering the effect of cadence on the CRP, a more out of phase movement pattern was displayed during the 60 RPM trial for the HK coupling (recovery phase) and a more in phase motion was displayed during the 120 RPM trial for the KA coupling (propulsive phases). Both these findings suggest the higher the cadence the more stable the resulting movement pattern. A stable coordinative pattern is able to be maintained despite perturbations to the system (Robertson, 2001) and according to Zanone et al. (2003), the more stable a movement pattern is, the lower the metabolic cost required to maintain the pattern at a given level of stability. This suggests that the coordination patterns exhibited at the higher cadences are more economical. This support for the use of a higher cadence is in agreement with Lucia et al. (2004) who found that for a fixed work rate, economy improves at increasing pedalling cadences and this improvement was attributed to a lower motor unit recruitment. The higher CRPv in the 60 RPM trial for the KA coupling during the recovery phase suggests a less consistent movement pattern and according to van Emmerick and van Wegen (2000) this is a sign of a less stable system. This is consistent with the CRP findings and also suggests that the variability present in the system is not beneficial to performance, something which has previously been suggested by Chapman et al. (2009). The fact that no differences in coupling motion were identified between work rates may be surprising given the significant differences between cadences and the interdependent relationship of work rate and cadence. However, the work rates investigated in this study were limited and greater ranges may be required to identify any differences which exist.

## CONCLUSION:

The results of this study suggest that changes in cadence may result in changes in stability and subsequently the economy for a given coordination pattern. This may have implications for both training and competition. Specifically the results support the use of a higher cadence. In addition, the less stable pattern identified during the recovery phases potentially highlights the need for further consideration of this phase by coaches. This study has been limited to intra-limb coordination however future work investigating interlimb coordination is advocated.

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# Kinematic changes during learning the longswing on high bar


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## INTRODUCTION:

Previous literature has reported differences between novice and expert technique during gross complex motor skills (Delignières et al., 1998), however, few have considered the nature of how novices' technique changes during a period of learning. Understanding how technique develops during learning provides precise information that can be used to influence feedback.

In men's gymnastics the longswing on high bar is a key skill which underpins the development of more complex skills. The biomechanics of performing successful longswings are well understood. Research has emphasised the importance of movements at the hip and shoulders, specifically, a hyper-extension to flexion action of the hips and hyper flexion to extension action of the shoulders occurring beneath the lower vertical (Arampatzis & Brüggemann, 1999; Yeadon & Hiley, 2000; Irwin & Kerwin, 2005). Irwin & Kerwin (2007) termed these movements Functional Phases (FP) since 70% of the gymnast's musculoskeletal work was found to occur during this part of the skill. Knowledge of the biomechanics of successful longswings, specifically the FPs, provides a theoretical underpinning to address more applied issues associated with learning longswing technique. Initial insights into kinematics associated with learning the longswing have been provided by Busquets et al. (2009) who described changes in coordination between the hip and shoulder for a novice cohort after a two month practice period in comparison to that of experienced gymnasts, and Williams et al. (2009) who provided a comparison of FP characteristics between performers at different stages of learning. However, the specifics of how longswing technique changes during a period of learning are not known. The purpose of the current study was to investigate the position of FPs of the 'looped bar longswing' performed by novices over a period of learning, in order to identify key kinematics of technique associated with learning.

***Fig. 1.*** *A full looped bar longswing and A participant attempting a looped bar longswing during a data collection session*

## METHOD:

Subjects:14 male participants with no prior high bar experience (age 20 ± 3 years, mass 73 ± 7 kg, height 1.76 ± 0.06 m), volunteered to take part in this study. The participants consented to learn the 'looped bar longswing' (LLS), a mechanically similar but safer variant of the traditional 'chalked bar longswing' (Irwin & Kerwin, 2005). The longitudinal study comprised an initial testing session in which participants were shown videos and received an explanation of the aims of the LLS before attempting the skill. Testing sessions required each participant to perform 5 trials of 3 swings with the ongoing aim of increasing swing amplitude. A gymnastics coach provided support to assist each participant in gaining initial angular momentum. Data were collected during each trial for each performer. The testing sessions were interspersed with training sessions throughout the study. During training sessions, longswing specific skills and conditioning exercises reflective of those used in contemporary coaching environment were performed in a gymnasium.

**Data Collection:** Unilateral kinematic data were collected using an automated 3D motion capture system (CODA) sampling at 200 Hz. Two CX1 CODA scanners (Charnwood Dynamics Ltd, UK) provided a field of view exceeding 2.5 m around the centre of the bar. Active markers were placed on the lateral aspect of each participant's right side at the estimated centre of rotation of the shoulder and the elbow, mid forearm, greater trochanter femoral condyle, lateral malleolus, fifth metatarsophalageal and the centre of the underside of the bar. For individuals, measures of height and mass were obtained, digital images facilitated the calculation of all other anthropometric data for use with a geometric inertia model (Yeadon, 1990) to obtain individual-specific body segment inertia parameters.

**Data Processing:** Swing 2 in each trial was analysed, ensuring a full independent attempt was being performed. Circle angle ($\theta_C$) was defined by the mass centre to bar vector with respect to the horizontal. In order to provide inter-performer comparisons of swings, data were interpolated in 1 degree increments of rotation about the bar. Lines joining the shoulder centre, greater trochanter and femoral condyle markers defined the hip angle ($\theta_H$). Shoulder angle ($\theta_S$) was defined by the lines joining elbow, shoulder and greater trochanter markers. Hip and shoulder angles ($\theta_H$; $\theta_S$) were differentiated to create angular velocity ($\omega_H$; $\omega_S$) profiles. 2D coordinate data were processed with the kernel smooth function (MathCad14™) with the smoothing parameter set to s = 0.10.

**Data Analysis:** The performance measure; Swing amplitude ($\theta_{CA}$), was defined as the circle angle between maximum height of the mass centre on the downswing to maximum height on the upswing. FP analysis of the hips was described by the position of maximum hyper-extension ($\theta_{CH1}$) to flexion ($\theta_{CH2}$) in $\theta_C$, and shoulders by maximum hyper flexion ($\theta_{CS1}$) to extension ($\theta_{CS2}$) in $\theta_C$. Differences across testing sessions were quantified using repeated measures ANOVA. Statistical significance was set at $p < .05$. Maulchy's test was used to determine the sphericity assumption within the data; where sphericity was violated probability was corrected according to the Greenhouse-Geisser procedure. Post hoc comparisons were made on the resultant data. Bonferroni corrections were applied for multiple comparisons.

## RESULTS:

Group mean $\theta_{CA}$ showed significant increases between session 1 (178 ± 42°) and session 8 (322 ± 41°), $p < .05$. The most rapid improvements occurred between the first and second testing sessions $p < .05$ (Fig. 1.).

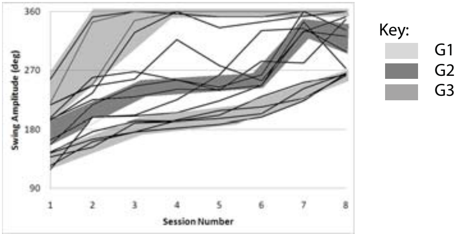


*Fig. 1. Mean swing amplitude over practice sessions.*

Increases in $\theta_{CA}$ followed there distinct trends (Fig. 1). Based on these trends, three groups were defined; G1, G2, G3 as follows:

**G1 *High learning rate*** (n=4): participants were able to perform the skill by session 3. Largest amplitude swings were demonstrated by these participants. (Fig. 1). FP analysis identified that $\theta_{CH1}$ became progressively later during the learning period ($p< .05$). During successful LLS in sessions 3-8, $\theta_{CS1}$ occurred earlier and θCS2 later than in sessions 1 and 2 (Table 1, Fig. 2).

**G2 *Variable learning rate*** (n = 5): demonstrated an initial increase in $\theta_{CA}$ between weeks 1 and 3, plateaued within the middle weeks before increasing during the penultimate week (Fig. 1). There were significant changes in hip FP variables $\theta_{CH1}$ and $\theta_{CH2}$ ($p< .05$), but not in the corresponding shoulder variables, $\theta_{CS1}$ and $\theta_{CS2}$.

**G3 *Low learning rate*** (n = 4): participants began and ended with the smallest swing amplitude but demonstrated a steady increase over the 8 sessions (Fig. 1). No significant differences occurred in $\theta_{CH1}$ during the 8 weeks, but $\theta_{CH2}$ advanced between sessions 1 and 8 ($p< .05$). After session 2, the onset of the shoulder FP ($\theta_{CS1}$) did not change (Fig. 2).

**Table 1. Session mean results for representative participants from G1, G2, G3 of the onset and termination of functional phase of the hips ($\theta_{CH1}$, $\theta_{CH2}$) and shoulders ($\theta_{CS1}$, $\theta_{CS2}$)**

| Session | G1 | | | | G2 | | | | G3 | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | CH1 | CH2 | CS1 | CS2 | CH1 | CH2 | CS1 | CS2 | CH1 | CH2 | CS1 | CS2 |
| 1 | 185 (8) | 290 (17) | 186 (10) | 368 (12) | 230 (12) | 325 (11) | 241 (16) | 346 (15) | 202 (10) | 288 (12) | 285 (9) | 320 (4) |
| 2 | 177 (10) | 278 (12) | 182 (6) | 355 (22) | 214 (3) | 332 (13) | 206 (11) | 371 (14) | 209 (3) | 288 (4) | 209 (10) | 322 (20) |
| 3 | 222 (60) | 338 (23) | 150 (8) | 381 (8) | 191 (4) | 313 (18) | 181 (11) | 385 (10) | 204 (4) | 297 (5) | 202 (4) | 335 (14) |
| 4 | 236 (38) | 345 (4) | 151 (11) | 384 (7) | 192 (6) | 321 (23) | 194 (31) | 343 (56) | 211 (6) | 304 (10) | 209 (6) | 358 (8) |
| 5 | 251 (9) | 350 (4) | 163 (10) | 392 (8) | 191 (9) | 298 (22) | 175 (13) | 346 (56) | 206 (4) | 296 (7) | 202 (4) | 354 (8) |
| 6 | 260 (12) | 348 (9) | 149 (3) | 387 (8) | 192 (7) | 335 (23) | 171 (17) | 393 (10) | 200 (18) | 300 (8) | 197 (18) | 352 (8) |
| 7 | 260 (11) | 356 (5) | 166 (4) | 398 (7) | 156 (31) | 349 (20) | 227 (68) | 403 (17) | 190 (16) | 308 (17) | 202 (33) | 361 (8) |
| 8 | 250 (13) | 351 (1) | 150 (13) | 392 (14) | 203 (11) | 346 (18) | 211 (44) | 330 (55) | 207 (5) | 335 (8) | 200 (31) | 371 (9) |

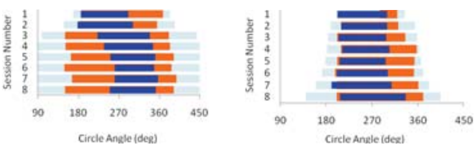


*Fig. 2. Mean functional phases (FP) of the hips (blue line) and shoulder (orange line) during the circle angle (light blue) for G1 (left) and G3 (right)*

## DISCUSSION:

The aim of this study was to describe changes in the position of FPs of the LLS performed by a group of novices during a period of learning. Swing amplitude ($\theta_{CA}$) significantly increased during the 8 week learning period. Three patterns of change in $\theta_{CA}$ (Group 1-3) were evident based on 2 criteria; the initial $\theta_{CA}$ and subsequent pattern change. Specifically, those participants who obtained the greatest $\theta_{CA}$ during session 1 were able to successfully perform the LLS by session 3 (G1). Four participants who had a mid-range $\theta_{CA}$ showed inconsistent increases in $\theta_{CA}$ over sessions (G2). Those who began with the smallest $\theta_{CA}$ in session 1 had the smallest $\theta_{CA}$ during session 8, sustaining a steady increase in $\theta_{CA}$ over sessions (G3). Therefore, the nature of improvement in the current skill appears to be closely related to skill level during initial attempts. Evidence from contemporary motor learning literature indicates that learning rate is individual and task specific even when a persistent change is apparent across subjects (Newell et al., 2001). During learning a lateral swinging task on a suspended platform, Teulier & Delignières (2007) found between participant differences in initial coordination pattern complexity, however during subsequent trials it was reported that technique evolved in a similar manner. Though, during this study initial $\theta_{CA}$ appears to be a good predictor of final success of the LLS. For similar tasks (Teulier & Delignières, 2007) it has been suggested that, based on Newell's (1986) categories of 'individual-specific organismic constraints' differences in initial skill levels can be related to organismic factors, as task and environmental constraints were constant for all participants.

FP analysis revealed individual specific changes throughout the learning period; analysis of a representative performer from G1, G2 and G3 were presented. Results showed differences in the ability to adjust the placement of the FP within the θC across sessions. Specifically, G1, appeared able to significantly change the onset and termination of the hip and shoulder FP throughout learning. G3 appeared to adjust hip FP, but not shoulder FP. In contrast, G2, were unable to significantly change the onset of the FP after session 2. Newell et al. (1989) suggested that variability in movement patterns permits the exploration of a motor-perceptual workspace, and was therefore an inherent characteristic of functional dynamical systems when learning a given motor task. For example, the more successful participants appear to be able to alter placement of both the hip and shoulder FPs in order to create a movement pattern that enabled them to increase $\theta_{CA}$, however it appeared that less successful participants were unable to alter these aspects of FPs in order to improve performance. Based on theories of motor learning, these findings have potential implications for the types of training intervention provided to a novice performer learning the LLS. For example, it could be interpreted that differences in initial $\theta_{CA}$ are related to organismic constraints of the system, where it is the ability to vary the onset and termination of the hip and shoulder FP which enable a progression to successful LLS. As such, it could be suggested that for a less successful performer (G3), skill progressions which promote actions temporally similar to that of the LLS for which firstly the hips (as per G2), and then the hips and shoulder actions (as per G1) would aid a performer in changing from an initially inadequate motor behaviour (Newell et al., 2001).

## CONCLUSION:

This study highlights the importance of quantifying changes in technique throughout learning, and on an intra-individual basis when seeking to investigate the nature of technique modifications. These experimental findings could provide key information which is required when considering motor learning in a sports context, in order provide the most effective feedback to a performer during skill acquisition. Additional work is required to explore if a relationship exists between the placement of the FP and $\theta_{CA}$ via kinetics analysis.

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