



HAL
open science

Voluntary control of pelvic frontal rotations in belly dance experts

Anne Tournillon, Isabelle A. Siegler

► **To cite this version:**

Anne Tournillon, Isabelle A. Siegler. Voluntary control of pelvic frontal rotations in belly dance experts. *Human Movement Science*, 2021, 77, pp.102791. 10.1016/j.humov.2021.102791 . hal-04289528

HAL Id: hal-04289528

<https://hal.science/hal-04289528>

Submitted on 1 Dec 2023

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16

Title Page

Voluntary control of pelvic frontal rotations in oriental dance experts

Anne Tournillon^{1,2}, Isabelle A. Siegler^{1,2}

Authors' affiliations:

1. Université Paris-Saclay, CIAMS, 91405, Orsay, France

2. Université d'Orléans, CIAMS, 45067, Orléans, France

Corresponding author:

Anne Tournillon annetournillon@gmail.com

Declarations of interest: none

17 **Abstract**

18 The present study investigated how oriental dance experts perform the “hip shimmy”, a complex
19 rhythmic dance movement consisting in a voluntary oscillation of the pelvis exclusively in the frontal
20 plane with maximised amplitude, with no movement of the upper trunk. The aims of this study were to
21 1) assess whether the amplitude and stability of the pelvic movement can be maximised in certain
22 postural and frequency conditions; and 2) investigate in a 1 to 3 Hz range whether it is indeed possible
23 to oscillate the pelvis only in the frontal plane and to dissociate this one-axis pelvic rotation from
24 potential spontaneous upper-trunk oscillations. Nineteen oriental dance experts performed this task in
25 three frequencies and three knee bending postures. Eight joint angles were calculated using the
26 kinematic data of 20 markers over the entire body collected with a motion capture system. Mean
27 amplitude, frequency, and spatial and temporal variability of frontal pelvic oscillations were analysed
28 to characterise motor performance and movement stability. Five Continuous Relative Phases (CRP)
29 were computed to identify the modes and stability of coordination patterns. The results showed that a
30 low posture enhances amplitude performance and that the pelvic oscillation amplitude tended to
31 decrease at 3 Hz, although between-condition differences remained small. Temporal stability was
32 highest at 2 Hz and significant inter-individual differences emerged at 3 Hz. CRP analysis revealed an
33 unpreventable coupling between pelvis and upper-trunk oscillations in the frontal and transversal
34 planes. A consistent antiphase coordination between transversal pelvis and upper-trunk may have been
35 caused by anatomical and counter-balancing constraints. In the frontal plane, multiple stable pelvis-
36 upper trunk patterns including inphase, out-of-phase and antiphase evolved to antiphase predominance
37 and inphase disappearance upon reaching 3 Hz. In sum, increasing frequency highlighted the
38 concomitance of two control phenomenon: the inter-individual differentiation in performance and
39 standardisation of the possible pelvis-upper-trunk patterns.

40 **Keywords**

41 Oriental dance; belly dance; rhythmic coordination; voluntary pelvic movement; motor performance;
42 continuous relative phase

43

44 **Highlights**

- 45 ▪ Dancers are able to separate the amplitude control of pelvic and upper-trunk oscillations.
- 46 ▪ Frontal and transversal pelvic rotations are coupled with anatomical constraints.
- 47 ▪ Inter-individual differences emerge when performing hip shimmy at 3 Hz.
- 48 ▪ There is a consistent antiphase pattern between the transversal pelvis and upper-trunk.
- 49 ▪ Coordination patterns between the frontal pelvis and upper trunk are frequency dependent.

50

52 1. Introduction

53 According to the non-linear dynamical system approach, coordination can be defined as a self-
54 organised intersegmental coupling spontaneously converging towards a consistent stable pattern, also
55 called state attractor. Several state attractors can coexist, resulting in a multistable system. Pioneering
56 work used the bimanual index finger oscillation rhythmic task to highlight the existence of two state
57 attractors, antiphase and inphase coordination patterns, whose transition from the first to the second
58 occurs abruptly along with increased variability when movement frequency is increased (Haken,
59 Kelso, & Bunz, 1985). The Continuous Relative Phase (CRP) between two body segments, which is
60 the difference between their phase angles at each moment of the movement, has been efficiently used
61 as an order parameter to assess such intersegmental coordination, with mean value and standard-
62 deviation of the CRP associated to the coordination mode and its stability, respectively (Burgess-
63 Limerick et al., 1993; Haken et al., 1985; J. A. S. Kelso et al., 1986; Van Emmerik & Wagenaar,
64 1996). Previous research used the CRP to examine the spontaneous coordination patterns between
65 pelvic and upper-trunk rotations during healthy and pathological human gait. Several studies showed
66 that with increasing gait velocity, a healthy coordination pattern between the pelvis and the upper
67 trunk in the transverse plane changed from a more inphase pattern to a more antiphase pattern,
68 accompanied by an increasing variability of the CRP during a “gradual” transition (Lamoth et al.,
69 2002; Van Emmerik & Wagenaar, 1996), while Low Back Pain or Parkinson subjects experienced
70 more difficulty achieving the antiphase pattern (Seay et al., 2011). The authors suggest that this
71 coupling provides dynamic stability and balance, and prevents injury: as velocity increases, pelvic
72 swing in axial rotation increases to help lengthen the stride, forcing the trunk to counterbalance the
73 pelvis in an antiphase pattern. Similarly, the literature often emphasises the lumbo-pelvic-hip
74 complex in stabilizing the centre of mass and the spine in order to prevent injury and maintain postural
75 balance (Chang et al., 2017; Tojima et al., 2016). However, very few studies investigate the control
76 and coordination mechanisms in voluntary pelvic movements, such as those performed in dance.

77 Oriental dance is an artistic and physical activity that requires complex, expert coordination. Its main
78 features can be characterised by wide and isolated movements of different parts of the trunk (pelvis,
79 chest, belly and shoulders) at different controllable speeds and amplitudes, possibly resulting from
80 underlying control mechanisms that are not observable in more common pedestrian tasks. In oriental
81 dance, the voluntary, rhythmic and isolated (dissociated from the rest of the body) pelvic rotation in
82 the frontal plane is called “hip lift”, or “pelvis side bending” when performed at slow or moderate
83 speed, and “hip shimmy” when performed at high speed. It is a rotation of the pelvis around the
84 anteroposterior axis alternating the clockwise and counter-clockwise directions, together with the
85 antiphase coordination of the lower limbs, i.e., alternating knee flexion and extension. This movement

86 is said to be expert when the speed and amplitude of the pelvis are controlled and when the three
87 following task constraints are respected: an almost immobile upper trunk (from the rib cage to the
88 shoulder line), continuous contact between the soles of the feet and the ground and exclusively frontal
89 pelvic rotation with equal amplitude on both sides. In oriental dance performance, efficient hip
90 shimmy execution is defined by the artistic relevance of the pelvis speed and amplitude choice, in
91 order to perfectly synchronise the movement with the rhythm and intensity of the music, in accordance
92 with the aforementioned constraints that define the task. Therefore, stability in controlling the
93 movement's amplitude and frequency is a key feature of expert performance. A recent study
94 emphasised the existence of several coordination patterns of the lumbar *erector spinae* muscles in the
95 control of this hip shimmy movement, depending on the movement frequency and the level of
96 expertise (Nugent & Milner, 2017). The authors identified three different *erector spinae* (ES) muscle
97 coordination patterns from the fourth lumbar vertebral (L4) to the tenth thoracic vertebral level (T10)
98 during the hip shimmy. An ipsilateral simultaneous pattern (S) of ES activation, causing an antiphase
99 relation between the first thoracic level (TH1) and the pelvic segment (PELV), was mainly observed in
100 novices. A diagonal synergy pattern (D) characterized by an antiphase muscle activation of the
101 ipsilateral ES between T10 and L4, resulting in the absence of a clear phase relationship between TH1
102 and PELV with a small amplitude of TH1, dominated in experts in 2 Hz and 3 Hz frequencies. An
103 asynchronous pattern (A) emerged in novices and experts from 3 Hz and above, and represented a
104 continuous phase shift of the ipsilateral ES activation and between the rotations of the different trunk
105 segments, in a travelling wave along the rostral-caudal direction, analogous to the EMG pattern
106 recorded in the swimming salamander. However, coordination and control mechanisms involved in
107 the maximisation of pelvis amplitude in the frontal plane have yet to be examined.

108 The aim of the present study was to use the hip shimmy movement performed by oriental dance
109 experts to investigate voluntary pelvic rotations and trunk coordination features that would not be
110 observable in more common gait gestures. We particularly aimed to 1) assess whether pelvic
111 movement amplitude and stability could be maximised in some postural and frequency conditions, and
112 2) characterise how the expected independent control of frontal pelvic from other trunk motions was
113 frequency dependent, using CRP analysis.

114

115 2. Methods

116 2.1 Participants

117 Nineteen female oriental dance experts volunteered to take part in this study (age: 30.3 ± 6.9 years,
118 height: 165.0 ± 0.1 cm, weight: 59.5 ± 7.0 kg, body mass index: 21.9 ± 2.3 , experience: 11.1 ± 5.1
119 years). They were recruited via social media and by e-mail exchanges over a one-month period. The
120 inclusion criteria were a minimum of three years of experience in oriental dance and a body mass
121 index less than or equal to 25. The participants read an information notice carefully and completed an
122 informed consent form approved by the Ethic Committee for Research (CER) of the University of
123 Paris-Saclay.

124 2.2 Experimental setup

125 The body movements were recorded by the QUALISYS optoelectronic system, using eight infrared
126 cameras with a sampling frequency of 179 Hz. Twenty reflective markers were placed on the
127 participants with double-sided tape according to the Plug-in Gait Full Body model as follows: over the
128 second metatarsal head (LTOE, RTOE); on the lateral malleolus (LANK, RANK); on the calcaneus of
129 each foot (LHEE, RHEE); over the upper lateral third of the right shank (RTIB); over the lower lateral
130 third of the left shank (LTIB); on the flexion/extension axis of the lateral side of both knees (LKNE,
131 RKNE); over the upper lateral third of the right thigh (RTHI); over the lower lateral third of the left
132 thigh (LTHI); over the left and right anterior superior iliac spine (LASI, RASI); over the left and right
133 posterior superior iliac spine (LPSI, RPSI); on the spinous process of the tenth thoracic vertebra (T10);
134 on the spinous process of the seventh cervical vertebra (C7); and on the left and right acromio-
135 clavicular joints (LSHO, RSHO).

136 The participants stood in front of a pole on which three visual markers were taped to help the dancers
137 adopt a standing posture with their knees flexed at an indicative angle. For each participant, the visual
138 markers were positioned at eye level for the three postures studied: a knee angle of 160 deg for High
139 Posture (HP); 145 deg for Intermediate Posture (IP), and 130 deg for Low Posture (LP), respectively.
140 These values were chosen in order to guarantee task feasibility for all participants: no less than
141 130 deg to prevent the heels from leaving the floor and balance loss, and no greater than 160 deg to
142 prevent the participants from locking their knees. The knee bending angle was measured with a
143 medical goniometer on the lateral side of the right knee, while the participants bent both knees equally.
144 Nevertheless, the achievement of the accurate instruction value was not required: no feedback on the
145 postural angle was given to the participants during the experiment.

146 2.3 Experimental procedure

147 The participants performed a three-minute standardised dance warm-up at the beginning of the
148 experiment. A 15-second static trial was recorded for each participant. In three posture (HP, IP, LP)

149 and three frequency (1 Hz, 2 Hz, 3 Hz) conditions, they were asked to perform a hip shimmy with the
150 greatest possible amplitude of pelvic rotation in the frontal plane, with equal pelvic inclination on both
151 sides. The three frequency conditions consisted of three frequencies led by a metronome (120 bpm,
152 240 bpm, 360 bpm) to which the participants were required to synchronise their pelvic motion. Two
153 15-second trials were performed in each condition. Conditions were organised in a pseudo-random
154 order by frequency: for each randomly-ordered frequency, the participants had to perform the
155 movement in each posture condition in a randomised order. Each trial was separated by a 10-second
156 break.

157 **2.4 Data processing**

158 One participant was removed from data analysis due to missing data during the experiment. The first
159 two seconds of each trial were discarded from data analysis. With the use of Visual3D (C-Motion,
160 Inc., Rockville, MD), the time series of six joint angles were calculated: rotation of the pelvis in the
161 frontal plane (*FronPel*), in the transverse plane (*TranPel*) and in the sagittal plane (*SagiPel*), and
162 rotation of the upper-trunk in the frontal plane (*FronSho*), in the transverse plane (*TranSho*) and in the
163 sagittal plane (*SagiSho*). For each participant, a kinematic model based on the static trial was
164 constructed with two rigid segments: pelvis and upper trunk. Both segments were associated with a
165 local segment coordinate system consistent with the global coordinate system (X as the medio-lateral
166 axis, Y as the antero-posterior axis, Z as the axial up axis). CODA Pelvis was used to construct the
167 pelvis segment and calculate the hip joint centres, from the LASI, RASI, LPSI and RPSI markers (Bell
168 et al., 1989, 1990). The local coordinate system of the pelvis (respectively upper trunk) segment was
169 defined with the X-axis passing through LASI and RASI (respectively LSHO and RSHO) markers, Y-
170 axis orthogonal to X-axis and included in the plane formed by the LASI, RASI and middle of LPSI
171 and RPSI (respectively LSHO, RSHO and C7) markers, and the Z-axis perpendicular to Y and X.
172 Using this kinematic model, we calculated the six pelvic and upper-trunk variables according to the
173 Visual3D default Cardan sequence X-Y-Z (Cole et al., 1993). Rotations of the local pelvis
174 (respectively upper-trunk) coordinate system in relation to the X, Y, Z-axis of the global coordinate
175 system corresponds to the sagittal, frontal and transversal pelvis (respectively upper-trunk) angles. We
176 chose to calculate the flexion angle of both knees (*LeftKne*, *RighKne*) as the convex 3D angle between
177 LTIB, LKNE and LTHI (respectively RTIB, RKNE and RTHI), where LKNE (respectively RKNE) is
178 the vertex of the angle. The angle-time series were filtered using a fourth order Butterworth low-pass
179 filter with a 12 Hz cut-off frequency.

180 The first ten cycles of each trial were used to perform the following analyses. For each cycle in a trial,
181 a custom-made MATLAB program (MATLAB r2014a, The MathWorks, USA) was used to identify the
182 minimum and maximum peak values of *FronPel*, subtracting each minimum value from the following
183 maximum value to obtain the amplitude of *FronPel*. The mean of the cycle amplitudes was computed

184 for each trial, as well as the standard deviation as an indicator of spatial stability. The mean and standard
185 deviation of the frontal pelvic oscillation frequency were computed using the time stamps of maximum
186 peak values of *FronPel*. The coefficient of variation of frequency, defined as the ratio of standard
187 deviation over mean frequency, was computed and used to evaluate the temporal stability of the pelvic
188 movement.

189 The mean range of motion of *FronSho*, *TranSho*, *SagiSho*, *TranPel* and *SagiPel* angles was computed
190 to assess the quasi-immobility of the upper-trunk and the predominance of frontal rotation of the pelvis
191 over its transversal and sagittal components in each trial. It was defined as the mean difference between
192 the maximum and minimum value of the angle across the ten shimmy cycles, defined between two
193 maximum peak values of *FronPel*.

194 A pelvic asymmetry index was computed to assess the participants' ability to rotate the pelvis with the
195 same amplitude to the left and right side during the trials. For both LASI and RASI markers, the mean
196 distances in the 3D space between their positions when the frontal pelvis rotation angle reached its
197 maximum and minimum peak values ($mean_{LASI}$ and $mean_{RASI}$) were calculated for each trial and used to
198 calculate $(mean_{LASI} - mean_{RASI}) / (mean_{LASI} + mean_{RASI})$. It varies between -1 and 1. The closer it is to 0,
199 the more symmetrical the pelvic rotation is on the left and right side.

200 In order to identify how the coordination patterns of the trunk were affected by frequency, we
201 calculated the continuous relative phase (CRP) between the following joint angles in each trial:
202 *FronPel* and *TranPel* ($\Delta\varphi_{FronPel-TranPel}$), *FronPel* and *SagiPel* ($\Delta\varphi_{FronPel-SagiPel}$), *FronPel* and
203 *FronSho* ($\Delta\varphi_{FronPel-FronSho}$), *FronPel* and *TranSho* ($\Delta\varphi_{FronPel-TranSho}$), and *FronPel* and *SagiSho*
204 ($\Delta\varphi_{FronPel-SagiSho}$). We calculated the angular velocity of each joint by differentiating each joint
205 angle. Then, the angular velocities and joint angles were normalised in order to range between -1 and 1
206 by rescaling to their respective minimum and maximum. The phase angle of each joint variable was
207 calculated as the inverse tangent of the division of the normalised velocity over the normalised joint
208 angle. The phase angle was calculated in the range from -180 to 180 deg. Finally, the CRP between
209 two variables is the result of subtracting the phase angles of these two variables. In order to restrain
210 CRP to the [-180; 180 deg] range, 360 deg were added or subtracted to the CRP values if needed.
211 For each trial, relative frequency histograms (proportions) of CRP values were computed to identify
212 different distribution patterns depending on the number of peaks: unimodal, multimodal or uniform
213 (no peak). When the histogram presented a clear unimodal distribution, the CRP was considered to be
214 relatively constant over the trial, so mean and standard deviation of the CRP were performed to assess
215 the mode and the stability of the coordination, using directional statistics (Burgess-Limerick et al.,
216 1991; Lamoth et al., 2002; Mello, 2005). We defined the coordination to follow an antiphase pattern if
217 the mean CRP within a trial ranged from ± 140 deg to ± 180 deg, out-of-phase pattern from ± 40 deg to
218 ± 140 deg, and inphase pattern from 0 deg to ± 40 deg.

219 **2.5 Statistical analysis**

220 The following analyses were performed using Matlab, Statistica 7.1 package (Statsoft, 2005) and
221 JASP (JASP Team, 2020). Only the data from the second trial in each condition were analysed, as the
222 first trial was considered a practice trial. The level of significance was set to $p = 0.05$. Normality was
223 verified with the Shapiro-Wilk test. BoxCox transformations were performed on a dataset when
224 normality was not met (Osborne, 2010; Sakia, 1992).

225 Paired sample *t*-tests performed on the mean of *RighKne* and *LeftKne* were used to verify whether
226 *RighKne* in the three posture conditions differed.

227 Repeated-measures ANOVAs were conducted to study the combined influence of the frequency (1 Hz,
228 2 Hz, 3 Hz) and the posture (LP, IP, HP) conditions on the dependent variables characterizing mean
229 performance and its variability, with a Bonferroni correction (Cramer et al., 2016). K-means clustering
230 and Mixed ANOVAs were also performed to test inter-individual differences (Macqueen, 1967).

231

232 **3. Results**

233 Figure 1 shows typical behaviour of joint angles during 12 cycles (five seconds) of a hip shimmy trial:
234 *FronPel*, *RighKne*, *LeftKne* and *TranPel* trace a clear oscillatory movement whereas *FronSho*,
235 *TranSho*, *SagiSho*, *SagiPel* show very small oscillations.

236 ***** Insert Figure 1 approximately here *****

237

238 **3.1 Motor performance**

239 **3.1.1 Compliance with task instructions**

240 In all trials of each participant, the mean knee angle respected an increasing order between LP, IP, and
241 HP conditions. Averaged across all participants, it reached 151.00 ± 5.19 deg in the HP condition,
242 147.49 ± 5.86 deg in the IP condition, and 141.54 ± 6.83 deg in the LP condition. The non-significant
243 Shapiro-Wilk test confirmed the normality assumption prior to performing *t*-tests on this postural
244 angle. Paired sample *t*-tests showed that the performed angles differed from each other, with a
245 significant difference between LP and IP ($t(53)=-13.75$; $p<0.001$), IP and HP ($t(53)=-8.47$; $p<0.001$),
246 LP and HP ($t(53)=-16.74$; $p<0.001$).

247 The mean range of motion of the upper trunk in the three planes (*FronSho*, *TranSho*, *SagiSho*) and
248 pelvic motion in the transversal and sagittal planes (*TranPel*, *SagiPel*) were always very small

249 (< 5 deg for all these variables except *TranPel* < 11 deg), as required in the task instructions (see Table
 250 1).

251

252 ***** Insert Table 1 approximately here *****

253

254 Table 1. Mean range of motion and standard deviation (degrees) of *FronSho*, *TranSho*, *SagiSho*,
 255 *TranPel* and, *SagiPel* in the nine experimental conditions, averaged between all participants.

	1 Hz			2 Hz			3 Hz		
	LP	IP	HP	LP	IP	HP	LP	IP	HP
<i>FronSho</i>	2.61±1.47	2.61±1.43	2.38±1.18	2.02±1.36	1.84±1.18	2.75±2.32	2.03±1.36	1.55±0.8	1.71±0.9
<i>TranSho</i>	4.58±2.41	4.78±2.79	4.26±2.17	3.64±2.42	3.16±2.06	4.39±2.90	3.26±1.97	2.95±2.56	3.36±2.37
<i>SagiSho</i>	2.67±0.88	2.79±1.06	2.75±1.07	1.93±0.60	1.84±0.54	2.44±0.85	2.07±0.70	1.66±0.49	1.83±0.52
<i>TranPel</i>	8.92±4.02	9.35±4.74	9.02±3.84	9.65±3.75	10.51±3.95	10.46±4.65	10.08±4.22	8.64±3.52	7.98±3.59
<i>SagiPel</i>	4.61±1.59	4.56±1.57	4.69±2.01	4.01±1.12	4.63±1.55	3.76±1.45	3.94±1.50	3.26±1.16	3.14±1.27

256

257 Finally, in all of the recorded trials, the absolute value of the asymmetry index ranged between 0.0006
 258 and 0.2, and the mean value was 0.07 ±0.05, showing good symmetry in pelvic oscillations. A
 259 repeated-measures ANOVA (3 frequencies × 3 postures) showed no significant influence of frequency
 260 and posture on the asymmetry index.

261

262 3.1.2 Pelvic oscillations in the frontal plane: Mean performance and variability

263 *Frequency*

264 The mean frequency of frontal pelvic oscillation was 1.02 ±0.003 Hz, 2.02 ±0.03 and 3.07 ±0.14 Hz in
 265 the 1, 2 and 3 Hz conditions, respectively, showing that the imposed frequency was respected.
 266 However, observations of large inter-individual differences in the 3 Hz condition led us to use K-
 267 means clustering in order to divide the participants into three different groups (Figure 2). Four
 268 participants performed a lower than expected frequency (F3- group), five participants a higher one
 269 (F3+ group) and nine participants performed the actual 3 Hz frequency (F3 group). One-sample *t*-tests
 270 showed that F3- ($t(11)=-19.94$; $p<0.001$) and F3+ ($t(14)=-9.24$; $p<0.001$) differed significantly from
 271 the instructed 3 Hz frequency, while F3 was not different from 3 Hz. Independent samples *t*-test

272 showed that in the 3 Hz condition, the three groups were significantly different from one another: F3-
 273 differed from F3+ ($t(25)=-13.94$; $p<0.001$), F3- differed from F3 ($t(37)=-7.41$; $p<0.001$) and F3+
 274 differed from F3 ($t(40)=9.22$; $p<0.001$).

275

276 ***** Insert Figure 2 approximately here *****

277

278 The following three ANOVAs were performed with an additional between-subject factor, resulting
 279 from this clustering.

280 The mixed ANOVA (3 frequencies \times 3 postures \times 3 groups) performed on the coefficient of variation
 281 of pelvic frequency yielded a significant effect of frequency condition ($F(2,30) = 24.68$; $p < 0.001$; η^2
 282 = 0.20). Post-hoc comparisons with the Bonferroni correction revealed that the coefficient of variation
 283 was significantly smaller at 2 Hz than at 1 Hz ($p<0.001$) and 3 Hz ($p<0.001$) (Figure 3).

284

285 ***** Insert Figure 3 approximately here *****

286 ***** Table 2 approximately here *****

287 Table 2. Mean pelvic amplitude (mean *FronPel*) in all participants (degrees) per condition.

Participant	1 Hz			2 Hz			3 Hz		
	LP	IP	HP	LP	IP	HP	LP	IP	HP
1	17.35	16.34	15.52	17.52	20.66	20.86	18.38	14.92	15.42
2	13.87	14.21	11.90	13.16	12.41	10.63	14.32	12.03	10.47
3	20.79	18.27	17.83	18.94	18.82	17.30	12.56	12.94	12.61
4	30.37	26.01	24.25	31.80	28.71	28.75	23.24	21.63	21.75
5	14.61	12.84	12.39	14.05	14.27	11.93	17.77	19.93	14.65
6	29.88	25.54	26.79	18.09	22.75	17.85	18.00	15.83	14.13
7	16.51	16.95	17.76	16.39	17.52	17.95	16.28	15.68	20.49
8	21.25	19.47	19.76	25.17	22.47	19.77	19.46	16.74	17.70
9	24.17	24.74	21.60	29.44	29.52	30.03	12.96	22.31	9.81
10	16.36	12.55	12.50	19.38	19.35	16.44	13.49	12.44	10.57
11	14.66	14.50	13.20	19.32	16.93	15.09	16.51	17.47	14.19
12	7.05	7.13	9.71	11.62	8.66	11.88	13.26	11.36	14.50
13	26.34	22.98	23.50	17.67	17.36	15.79	3.14	10.04	4.91
14	17.59	16.33	15.18	23.24	22.97	22.78	25.07	24.50	23.36
15	24.29	25.26	22.89	25.52	25.34	24.57	16.07	17.26	16.21
16	17.36	19.82	20.30	16.20	17.07	14.99	18.89	9.21	14.83
17	23.18	24.65	18.64	24.16	19.20	21.63	22.85	23.01	21.70

18 20.84 21.50 19.99 17.81 21.58 21.16 15.93 17.04 16.47

288

289 *Amplitude*

290 Mean amplitude (over a trial) of pelvic oscillation in the frontal plane ranged from 3.14 deg to
291 31.80 deg across participants (Table 2). A mixed ANOVA (3 frequencies \times 3 postures \times 3 groups)
292 performed on this variable yielded a significant main effect of posture ($F(2,30) = 5.60$; $p = 0.01$; $\eta^2 =$
293 0.01), of frequency ($F(2, 30) = 3.58$; $p = 0.04$; $\eta^2 = 0.05$), as well as a marginally significant
294 interaction between frequency and group ($F(4, 30) = 2.57$; $p = 0.06$; $\eta^2 = 0.08$) (Figure 4.a). Post-hoc
295 comparisons with the Bonferroni correction revealed that mean amplitude was significantly smaller in
296 the high posture than in the low posture (mean difference = 1.44 deg; $p = 0.007$). Concerning the
297 marginal interaction between frequency and group, Figure 4.b shows that the F3+ group exhibits the
298 lowest amplitude in the 3 Hz condition. A mixed ANOVA (3 frequencies \times 3 postures \times 3 groups)
299 performed on the standard deviation of the within-trial frontal pelvic amplitudes showed no significant
300 influence of any of the three factors, nor any significant interactions between them, suggesting a
301 relatively constant spatial variability across groups and conditions.

302 ***** Insert Figures 4.a and 4.b approximately here *****

303

304 **3.2 Trunk coordination mode and stability: CRP analysis**

305 The individual CRP proportion distributions were averaged for each frequency condition (Figure 5).
306 Three types of distribution were identified depending on the number of peaks. A unimodal distribution
307 reflected either a “more” inphase coordination mode with a peak around 0 deg, or antiphase
308 coordination mode with two peaks around ± 180 deg. A bimodal distribution reflected the coexistence
309 of two coordination modes among participants, inphase and antiphase with three peaks around
310 ± 180 deg and 0 deg. However, in both cases, non-null values between peaks show that some out-of-
311 phase patterns could be observed and analysed thereafter. Finally, a uniform distribution with no peak
312 reflected the absence of stable coordination.

313 ***** Insert Figure 5 approximately here *****

314

315 $\Delta\varphi_{FronPel-FronSho}$ displays a bimodal distribution with three peaks around 0 deg and ± 180 deg at
316 1 Hz and 2 Hz, whereas it shows a unimodal distribution with two peaks around ± 180 deg at 3 Hz. In
317 order to better understand these relatively complex distributions, mean and standard deviation of
318 $\Delta\varphi_{FronPel-FronSho}$ were calculated for each trial, as 93.8% of all individual trial distributions were
319 unimodal. Figure 6 shows the mean individual relative phase values for the three frequencies,

320 revealing the prevalence of the three coordination types as a function of frequency. At 1 Hz, 37% of
321 the trials exhibited an inphase pattern, 29.6% and out-of-phase pattern and 33% an antiphase pattern.
322 At 2 Hz, the figures were 22.2%, 35.2% and 42.6%, respectively. At 3 Hz, more than half of the trials
323 exhibited antiphase patterns (64.8%) and a very small proportion exhibited an inphase pattern (5.5%),
324 while 29.6% exhibited an out-of-phase pattern. An inphase pattern means that *FronSho* rotates with
325 the right shoulder downwards when *FronPel* rotates with the right hip downwards, while it rotates
326 with the left hip downwards in an antiphase pattern. An out-of-phase pattern means that *FronSho* and
327 *FronPel* minimum or maximum peak rotations do not coincide in time but are phase-shifted (Figure
328 7).

329 ***** Insert Figure 6 approximately here *****

330 ***** Insert Figure 7 approximately here *****

331

332 In order to see if the variability increases in out-of-phase patterns, standard deviation was plotted for
333 each trial (Figure 8). Standard deviation of each trial seemed more or less constant through the
334 patterns. The mean standard deviation was 55.72 ± 11.79 deg in the inphase trials, 63.41 ± 8.97 deg in
335 the out-of-phase trials, and 52.55 ± 12.63 deg in the antiphase trials.

336 $\Delta\varphi_{FronPel-Transho}$ displays a clear unimodal distribution around 0 deg regardless of the frequency
337 condition (Figure 5). More than 94% of all individual histograms showed this same unimodal
338 distribution pattern around 0 deg. Averaged between participants, the mean value of
339 $\Delta\varphi_{FronPel-Transho}$ was 24.43 ± 30.94 deg, and the mean standard deviation was 47 ± 13.73 deg. This
340 inphase pattern means that when *FronPel* rotates downward with the right hip, the *Transho* rotates
341 forward with the right shoulder.

342 Both $\Delta\varphi_{FronPel-SagiPel}$ and $\Delta\varphi_{FronPel-SagiSho}$ display a diffuse bimodal distribution around 0 deg
343 and ± 180 deg at 1 Hz and an uniform distribution at 2 Hz and 3 Hz. However, the individual histogram
344 displays no coordination pattern consistency between subjects, and none of them displayed a clear
345 unimodal distribution.

346 $\Delta\varphi_{FronPel-TranPel}$ depicts a clear antiphasic pattern in all frequency conditions (Figure 5), which is
347 observed in more than 96% of all individual histograms. This antiphase pattern means that when
348 *FronPel* rotates downward with the right hip, *TranPel* rotates backward with the right hip. Averaged
349 between participants, the absolute mean value of $\Delta\varphi_{FronPel-TranPel}$ reached 156.99 ± 30.72 deg, and
350 mean standard deviation reached 33.39 ± 16.07 deg.

351

352 **4. Discussion**

353 In the present study, we investigated the extent to which it is possible to intentionally oscillate the
354 pelvis in the frontal plane independently from rotations in the two other pelvic planes (i.e. transverse
355 and sagittal planes) while the upper trunk remains still, as expected in an oriental dance movement
356 called the “hip-shimmy”. More specifically, the first aim was to assess whether pelvic movement
357 amplitude and stability could be maximised in certain postural and frequency conditions in order to
358 contribute to performance optimisation in this well-known dance movement, whose kinematic
359 properties have rarely been documented. The second and more fundamental aim was to characterise
360 how the expected phase unlocking (or potential phaselocking) of 3D pelvis and upper-trunk rotations
361 was frequency dependent. The ability of the oriental dance experts to independently control the
362 different rotations of the pelvis, as well as the pelvis from upper-trunk movements was analysed with
363 continuous relative phases.

364 *Performance features.*

365 The mean amplitude of frontal pelvis rotation among experts oriental dancers ranged from 15 to
366 20 deg depending on the condition, which is higher than in gait: approximately 10 deg when walking
367 at preferential speed (Taylor et al., 2003) and 7 deg in racewalking (Majed et al., 2012). A main
368 significant effect of the role of posture was observed, with a low posture enhancing performance.
369 However, the size of this effect was small with a difference of less than 2 deg between the movement
370 amplitude in high and low postures. The anatomical constitution of the hips might explain this postural
371 effect. As posture lowers, the knees and hips are more flexed and the anterior ligaments of the hips are
372 looser. This increases the abduction and adduction amplitude ability of the hips, both of which occur
373 when the pelvis is rotated in the frontal plane (Calais-Germain, 2013).

374 Concerning the significant effect of movement frequency, mean performance was similar and highest
375 in the 1 and 2 Hz conditions. A drop of approximately 3-4 deg was observed at 3 Hz. In cyclical
376 movements, it is often observed that the amplitude of cycle drops inversely as frequency is increased
377 and that amplitude and frequency cannot necessarily be controlled independently (B. A. Kay, J. A.
378 Kelso, E. L. Saltzman, G Schöner, 1987). Finally, the absence of a significant interaction between
379 posture and frequency effects, as well as parallel curves in Figure 4.a, show that the frequency effect
380 was not posture dependent.

381 *Is there independent control of pelvic motion in the frontal plane?*

382 The analysis of mean joint motion amplitudes showed that expert oriental dancers are able to minimise
383 pelvic sagittal and upper-trunk motions which were much smaller (although not completely absent)
384 than pelvic motion in the frontal plane. It was less the case for the pelvic rotation in the transverse
385 plane, which reached approximately 11 deg in some conditions. These results appear to show that, at
386 least to some extent, expert oriental dancers are able to dissociate the pelvic motion from the upper-
387 trunk motion, as well as the pelvic motions around the three pelvic rotation axes. Yet, the analysis of

388 the continuous relative phases enables us to be more specific and moderate about this latter assertion.
389 Indeed, while the upper-trunk motion remained very small in all the conditions, the results showed
390 some clear inphase phase locking between the pelvic motion in the frontal plane and the shoulder
391 motion in the transverse plane, in all three frequency conditions. There was also a clear antiphase
392 phase locking between the pelvic motion in the frontal plane and the transverse planes for all
393 participants in all frequency conditions. By deduction, the transversal pelvis and upper trunk are
394 systematically coordinated in antiphase. These very consistent observations tend to show that upper-
395 trunk and pelvic motion in the transverse plane may be due to either mechanical, anatomical or both
396 factors that would be impossible to suppress even after more than three years of training. The
397 antiphase relationship between the pelvis and shoulders in the transverse plane could reflect the
398 counter-balancing mechanism often suggested in gait analysis (Lamoth et al., 2002; Van Emmerik &
399 Wagenaar, 1996) and/or that large voluntary pelvic movements induce small but systematic
400 mechanical upper-trunk vibrations.

401 Conversely, other relative phases exhibit frequency-dependent distributions. At 1 Hz, three
402 distributions demonstrate that more or less inphase, antiphase or even out-of-phase coordination
403 modes are possible, namely between the frontal pelvic motion and respectively sagittal pelvic motion,
404 frontal and sagittal upper-trunk motion. Unlike bimanual rhythmic tasks which present only two stable
405 states, either inphase or antiphase (J.A. Kelso, 1984), we found that coordination between the frontal
406 pelvis and the upper trunk can reach multiple stable levels. This finding is consistent with Van
407 Emmerik et al (1996) who showed that multiple stable coordination patterns exist in trunk motion
408 during gait when velocity is increased.

409 In our study, a reduction in the number of coordination modes or even no phase locking was observed
410 when frequency was increased. At 2 and 3 Hz, there was no stable phase relationship between frontal
411 pelvic oscillation and sagittal pelvic or shoulder oscillations. Thus, high frequency enhances some
412 dissociation between joint rotations expected in the hip-shimmy movement, whereas, as mentioned
413 above, other phase-locked coordination patterns seem impossible to prevent.

414 *Frequency-dependent variability in motor performance and inter-individual differences*

415 Frequency also has an influence on movement variability. The temporal variation of oscillations was
416 significantly smaller in 2 Hz than in 1 Hz and 3 Hz, which can be interpreted as greater stability at
417 2 Hz than at the other two studied frequencies. Two main interpretations are possible. The 2 Hz
418 frequency may be close to the natural frequency of the hip shimmy movement since it is known that
419 movement natural frequencies enhances stability (Jordan et al., 2007; Rosenblum et al., 1988). It could
420 also be that this frequency is the one most often practiced by our participants when moving in rhythm
421 with music. It could well be that pelvic movements performed at 1 Hz are more a succession of
422 discrete movements than a true rhythmic movement. It is known that rhythmic movement cannot be

423 produced when it is too slow (Park et al., 2017). Regarding 3 Hz, more inter-individual differences
424 arose. The 3 Hz condition seemed more difficult to maintain for half of our participants, who naturally
425 converged into either a higher or a lower frequency, and exhibited larger temporal variability. This
426 result is in accordance with Nugent et al. (2017) who mention that 3 Hz is the threshold frequency
427 where tempo would be difficult to maintain, leading to a either lower or higher frequency than
428 expected. In the non-linear dynamical systems view, we could interpret this frequency as being located
429 in an instability area between two stable attractors, yielding critical fluctuations in movement. Nugent
430 et al. (2017) found that a new pattern of *erector spinae* muscle activation emerges from 3 Hz and
431 could not be performed at lower frequencies (the “A” pattern). We believe that our expert participants
432 who performed at a higher frequency than the imposed 3 Hz may have switched to this A pattern. This
433 group also performed smaller pelvic oscillations at 3 Hz than the other participants.

434 The frequency threshold, above which dancers spontaneously shift to this A pattern, could be related
435 to the level of expertise and thus be used to characterise oriental dancers’ performance level. In future
436 studies, a between-group design experiment with groups of different levels of expertise and higher
437 frequencies tested (up to 6 Hz as suggested by Nugent et al (2017)) could provide more insight into
438 this issue of inter-individual differences and frequency-related variability.

439 Although not directly studied in the present paper, there were also some inter-individual differences in
440 the knee bending posture adopted by the participants. We contend that this is a factor of inter-
441 individual variability in mean performance (maximal pelvic frontal motion) since posture was shown
442 to have a significant (but small) main effect. Still, this effect did not interact significantly with the
443 frequency-dependent effects on performance that were observed and discussed.

444 **Conclusion**

445 Oriental dance provides an innovative approach for the study of human motor control and coordination
446 to provide insight into complex control mechanisms of pelvic motion. Although dancers were able to
447 maintain a wide, stable, voluntary pelvic rotation in the frontal plane while minimizing the other pelvis
448 and upper-trunk rotations, frontal pelvis control could not be kept perfectly independent because of
449 anatomical/mechanical constraints, yielding some unpreventable coordination in the trunk. Moreover,
450 frequency could most probably be a differentiating factor of the expertise level between individuals,
451 and a control parameter of the different pelvic coordination pattern occurrences. The ability to
452 voluntarily control the pelvic motion in a hip shimmy through changing and high frequencies among
453 dancers of different levels of expertise and novices should be explored in further studies.

454 **References**

- 455 Bell, A. L., Brand, R. A., & Pedersen, D. R. (1989). Prediction of hip joint centre location from
456 external landmarks. *Human Movement Science*, 8(1), 3-16. [https://doi.org/10.1016/0167-](https://doi.org/10.1016/0167-9457(89)90020-1)
457 [9457\(89\)90020-1](https://doi.org/10.1016/0167-9457(89)90020-1)
- 458 Bell, A. L., Pedersen, D. R., & Brand, R. A. (1990). A comparison of the accuracy of several hip
459 center location prediction methods. *Journal of Biomechanics*, 23(6), 617-621.
460 [https://doi.org/10.1016/0021-9290\(90\)90054-7](https://doi.org/10.1016/0021-9290(90)90054-7)
- 461 Burgess-Limerick, R., Abernethy, B., & Neal, R. J. (1991). Note: A Statistical Problem in Testing
462 Invariance of Movement Using the Phase Plane Model. *Journal of Motor Behavior*, 23(4),
463 301-303. <https://doi.org/10.1080/00222895.1991.9942041>
- 464 Burgess-Limerick, R., Abernethy, B., & Neal, R. J. (1993). Relative phase quantifies interjoint
465 coordination. *Journal of Biomechanics*, 26(1), 91-94. [https://doi.org/10.1016/0021-](https://doi.org/10.1016/0021-9290(93)90617-N)
466 [9290\(93\)90617-N](https://doi.org/10.1016/0021-9290(93)90617-N)
- 467 Calais-Germain, B. (2013). *Anatomie pour le mouvement, tome 1 : Introduction à l'analyse des*
468 *techniques corporelles (French Edition)* (5th éd.). Editions DésIris.
- 469 Chang, M., Slater, L., Corbett, R., Hart, J., & Hertel, J. (2017). Muscle activation patterns of the
470 lumbo-pelvic-hip complex during walking gait before and after exercise. *Gait & Posture*, 52,
471 15-21.
- 472 Cole, G. K., Nigg, B. M., Ronsky, J. L., & Yeadon, M. R. (1993). Application of the Joint Coordinate
473 System to Three-Dimensional Joint Attitude and Movement Representation: A
474 Standardisation Proposal. *Journal of Biomechanical Engineering*, 115(4A), 344-349.
475 <https://doi.org/10.1115/1.2895496>
- 476 Cramer, A. O. J., van Ravenzwaaij, D., Matzke, D., Steingroever, H., Wetzels, R., Grasman, R. P. P.
477 P., Waldorp, L. J., & Wagenmakers, E.-J. (2016). Hidden multiplicity in exploratory multiway
478 ANOVA: Prevalence and remedies. *Psychonomic Bulletin & Review*, 23(2), 640-647.
479 <https://doi.org/10.3758/s13423-015-0913-5>
- 480 Haken, H., Kelso, J. A. S., & Bunz, H. (1985). A Theoretical Model of Phase Transitions in Human
481 Hand Movements. *Biological Cybernetics*, 51, 347-356.

482 Jordan, K., Challis, J. H., & Newell, K. M. (2007). Walking speed influences on gait cycle variability.
483 *Gait & Posture*, 26(1), 128-134. <https://doi.org/10.1016/j.gaitpost.2006.08.010>

484 Kay, B. A., Kelso, J. A., Saltzman, E. L., & Schöner, G. (1987b). Space–time behavior of single and
485 bimanual rhythmical movements: Data and limit cycle model. *Journal of Experimental*
486 *Psychology: Human Perception and Performance*, 13(2), 178-192.
487 <https://doi.org/10.1037/0096-1523.13.2.178>

488 Kelso, J. A. (1984). Phase transitions and critical behavior in human bimanual coordination. *American*
489 *Journal of Physiology-Regulatory, Integrative and Comparative Physiology*, 246(6),
490 R1000-R1004. <https://doi.org/10.1152/ajpregu.1984.246.6.R1000>

491 Kelso, J. A. S., Scholz, J. P., & Schöner, G. (1986). Nonequilibrium phase transitions in coordinated
492 biological motion: Critical fluctuations. *Physics Letters A*, 118(6), 279-284.
493 [https://doi.org/10.1016/0375-9601\(86\)90359-2](https://doi.org/10.1016/0375-9601(86)90359-2)

494 Lamoth, C. J. C., Beek, P. J., & Meijer, O. G. (2002). Pelvis–thorax coordination in the transverse
495 plane during gait. *Gait & Posture*, 16(2), 101-114. [https://doi.org/10.1016/S0966-](https://doi.org/10.1016/S0966-6362(01)00146-1)
496 [6362\(01\)00146-1](https://doi.org/10.1016/S0966-6362(01)00146-1)

497 Macqueen, J. (1967). Some methods for classification and analysis of multivariate observations.
498 *Proceedings of the Fifth Berkeley Symposium on Mathematical Statistics and Probability*,
499 281-297.

500 Majed, L., Heugas, A. M., & Siegler, I. (2012). Learning an energy-demanding and biomechanically
501 constrained motor skill, racewalking: Movement reorganization and contribution of metabolic
502 efficiency and sensory information. *Human Movement Science*, 31, 1598-1614.

503 Mello, C. (2005). Statistiques circulaires et utilisations en psychologie. *Tutorials in Quantitative*
504 *Methods for Psychology*, 1(1), 11-17. <https://doi.org/10.20982/tqmp.01.1.p011>

505 Osborne, J. (2010). *Improving your data transformations: Applying the Box-Cox transformation*.
506 *15*(12), 10.

507 Park, S.-W., Marino, H., Charles, S. K., Sternad, D., & Hogan, N. (2017). Moving slowly is hard for
508 humans : limitations of dynamic primitives. *Journal of Neurophysiology*, 118(1), 69-83.
509 <https://doi.org/10.1152/jn.00643.2016>

510 Rosenblum, L. D., & Turvey, M. T. (1988). Maintenance tendency in co-ordinated rhythmic
511 movements : Relative fluctuations and phase. *Neuroscience*, 27(1), 289-300.
512 [https://doi.org/10.1016/0306-4522\(88\)90238-2](https://doi.org/10.1016/0306-4522(88)90238-2)

513 Sakia, R. M. (1992). The Box-Cox Transformation Technique: A Review. *The Statistician*, 41(2), 169.
514 <https://doi.org/10.2307/2348250>

515 Seay, J. F., Van Emmerik, R. E. A., & Hamill, J. (2011). Low back pain status affects pelvis-trunk
516 coordination and variability during walking and running. *Clinical Biomechanics*, 26(6),
517 572-578. <https://doi.org/10.1016/j.clinbiomech.2010.11.012>

518 Taylor, N. F., Evans, O. M., & Goldie, P. A. (2003). The effect of walking faster on people with acute
519 low back pain. *European Spine Journal*, 12(2), 166-172. [https://doi.org/10.1007/s00586-002-](https://doi.org/10.1007/s00586-002-0498-3)
520 0498-3

521 Tojima, M., Ogata, N., Inokuchi, H., & Haga, N. (2016). Three-dimensional motion analysis of
522 lumbopelvic rhythm during lateral trunk bending. *Journal of Physical Therapy Science*, 28(8),
523 2342-2346. <https://doi.org/10.1589/jpts.28.2342>

524 Van Emmerik, R. E. A., & Wagenaar, R. C. (1996). Effects of walking velocity on relative phase
525 dynamics in the trunk in human walking. *Journal of Biomechanics*, 29(9), 1175-1184.
526 [https://doi.org/10.1016/0021-9290\(95\)00128-X](https://doi.org/10.1016/0021-9290(95)00128-X)

527

528

529 **Legends**

530

531 Figure 1. Joint angles in five seconds of a hip shimmy trial of one participant in the 2 Hz and Low
532 Posture condition.

533

534 Figure 2. Mean frequencies (\pm standard error) of frontal pelvic oscillation (*FronPel*) as a function of
535 experimental conditions in three groups of participants (white circle: F3- group, grey square: F3, black
536 diamond: F3-).

537

538 Figure 3. Mean (\pm standard error) coefficient of variation of frontal pelvic frequency as a function of
539 metronome frequency in the F3- group (circle), F3 group (square), F3+ group (diamond) (prior to
540 BoxCox transformation).

541

542 Figure 4.a) Mean amplitude of pelvic motion in the frontal plane (*FronPel*) in all conditions as a
543 function of metronome frequency (white square: Low Posture; grey square: Out-of-phase Posture;
544 black square: High Posture). b) Mean amplitude of pelvic motion in the frontal plane for the three
545 groups as a function of metronome frequency.

546

547 Figure 5. Proportion (y-axis) distributions of CRP values in the [-180 ; 180] deg range (x-axis), across
548 frequency trials, averaged in all participants.

549

550 Figure 6. Mean $\Delta\varphi_{FronPel-FronSho}$ of each individual trial as a function of frequency condition (in
551 each frequency graph, from left to right, successive means of LP, IP, HP of each participant).

552

553 Figure 7. Example of Out-of-phase pattern coordination between *FronPel* and *FronSho* in participant
554 no. 2, in the 2 Hz and HP condition. Mean CRP: 103.73 ± 57.74 deg. Top: unimodal histogram of
555 CRP. Bottom: *FronPel* (black line) and *FronSho* (grey line).

556

557 Figure 8. Mean (circles) and standard deviation (error bars) of $\Delta\varphi_{FronPel-FronSho}$ of each trial, sorted
558 along the x-axis according to the increasing order of mean $\Delta\varphi_{FronPel-FronSho}$. Inphase pattern trials
559 (white circles), out-of-phase pattern trials (grey circles) and antiphase pattern trials (black circles).