Whole body coordination and knee movement control during five rehabilitation exercises.

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Abstract

Knee rehabilitation exercises to improve motor control, target movement fluency and displacement variability. Although knee movement in the frontal plane during exercise is routinely assessed in clinical practice, optimal knee control remains poorly understood. In this study, twenty-nine healthy participants (height: 1.73±0.11 m, mass: 73.5±16.4 kg, age: 28.0±6.9 years) performed four repetitions of five rehabilitation exercises whilst motion data were collected using the VICON PlugInGait full body marker set. Fluency and displacement variability were calculated for multiple landmarks, including Centre of Mass (CoM) and knee joint centres. Fluency was calculated as the inverse of the average number of times a landmark velocity in the frontal plane crossed zero. Variability was defined as the standard deviation of the frontal plane movement trajectories. CoM fluency and displacement variability were significantly different between tasks ($p<0.001$). CoM displacement variability was consistently smallest compared to the constituent landmarks ($p<0.005$). This was interpreted as a whole body strategy of compensatory variability constraining CoM frontal plane movement. Ipsilateral knee fluency ($p<0.01$) and displacement variability ($p<0.001$) differed substantially between tasks. The role of the weight-bearing knee seemed dependent on task constraints of the overall movement and balance, as well as constraints specific for knee joint stability.

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Rehabilitation following acute knee injury and/or surgery includes exercises to improve motor control to learn to cope with movement challenges encountered during everyday activities and sport. Rehabilitation protocols in published studies have focused on perturbation exercises, muscle strengthening, and neuromuscular control exercises or generalised programmes that combine these three exercise types [1-7]. In clinical practice, functional exercises can be defined as training of everyday movements that are most relevant to the individual and affected or put at risk by a particular condition. For example, poor knee control in the frontal plane during hopping has been associated with increased risk of anterior cruciate ligament (ACL) injury [8] and increased medial knee loading in the frontal plane during walking has been associated with development of osteoarthritis [9]. Therefore, knee movement assessment in the frontal plane is routine clinical practice [10] and improvement of knee control in the frontal plane is an important objective during functional exercises which will focus on neutral alignment of the knee [8, 10-11], movement fluency [12-14] and displacement variability [15].

Fluency or smoothness is a characteristic of natural movement [16] and was suggested to be an important coordinative control parameter in multi-joint movement [17]. This concept was, for instance, used to study fine motor control in patients with chronic wrist pain [12]. Fluency in studies of medio-lateral knee control [13-14] was defined as the number of times per second the velocity of the knee position in the frontal plane crossed zero calculated by a method adapted from Smeulders et al. [12]. Roos et al. [13] found that following landing from a hop knee frontal plane movement fluency was significantly different between three groups; knee movement was least fluent in patients with ACL deficiency due to injury (ACLD); patients with anterior cruciate ligament reconstruction (ACLR) were intermediate; and knee movement was most fluent in healthy control subjects. Panos et al. [18] found similar results when asking people with ACL injury to perform drop landing tasks. However, Button et al. [14] found that during walking knee fluency was significantly higher in the ACLD group; there were no significant differences between the ACLR and healthy groups. In the same study during single leg squatting (SLSQ) knee fluency was significantly less for the ACLR group compared to the other two; and during double leg squatting (DLSQ) there were no significant differences between the three groups. The diverse results for these functional movements may relate to the variety of motor solutions people typically use for each task (task dependence). This provides challenges for theoretical interpretation as well as for translation to clinical practice. Therefore further exploration of the role of knee movement during these functional exercises is required to improve our understanding how rehabilitation can be optimised.

Variability is equally considered part of normal movement and is considered to be essential for effective motor control and motor learning [19-20]; for instance better coordination has been associated with reduced magnitudes of movement fluctuations [21]. It is unlikely that minimizing frontal plane knee movement is principally the parameter that is controlled during normal functional movement. Rather, it has been proposed that an important movement goal during gait is to constrain the medio-lateral oscillation of the centre of mass (CoM) [22-24]. Bauby and Kuo [24] proposed that medio-lateral gait stability requires active feedback control by a compensatory use of mechanisms, such as lateral torso movement and ankle eversion/inversion activity. In our interpretation all limbs and the trunk would be key actors in such a complex system of motor control to constrain CoM frontal plane movement; therefore presumably knee activity of the stance leg is part of such a complex system. It has been suggested that movement of all components in the system to constrain CoM movement [25] is achieved through a whole body synergy whereby movement of all constituent parts are related to each other. Movement variability of the constituent parts to some extent can cancel each other out so that an overall control parameter moves with reduced variability [21]. This co-variation or compensatory variability [26] during whole body movement has been demonstrated in a number of studies involving healthy subjects [25, 27-28]. Inter-limb coordination as part of a complex system is controlled dynamically to create the desired movement patterns [24]. It is unlikely that a single optimal solution exists for a given task. Rather, intra and inter-individual differences in task solutions are considered characteristic of normal movement and useful for adaptive motor control [26] and therefore rehabilitation. These principles of motor control are pertinent to increase the understanding of a variety of movements and thus the rehabilitation process. We therefore explored the fluency and displacement variability in five functional whole body exercises relevant to rehabilitation. By exploring these parameters in a healthy group of people we aimed to generate a basis for interpreting changes observed in patients following knee injuries.

For the analysis we used a simplified eight component kinematic model representing the trunk and the four limbs to explore movement coordination in relation to CoM control. We expected that CoM fluency and displacement variability are task dependent; we therefore hypothesised there would be significant differences between tasks for these variables. We also expected that compensatory variability would result in constrained CoM movement and therefore hypothesised that fluency would be largest and displacement variability smallest for the CoM compared to the constituent components of the motor control system that is limbs and trunk. A larger number of movement corrections with larger displacement variability compared to the CoM we assumed would be consistent with the concept of compensatory variability. Specifically with respect to the role of the weight-bearing knee within this overall synergy we expected for each task to demonstrate smaller fluency and larger displacement variability compared to the CoM. These knee variables were also expected to be task-dependent. We therefore hypothesised there would be significant differences between tasks for these variables of the weight-bearing knee.

Methods

Twenty nine healthy subjects (demographics are in Table 1) participating in a larger cross-sectional study provided informed consent to participate in this study. Inclusion criteria were that subjects were aged between 18 and 50 years and healthy; they were required to have no condition or previous trauma that had required clinical intervention and/or no assistive devices to either limb including ankle, knee and hip. Ethical approval for this study was obtained from the South East Wales Research Ethics Committee (Reference: 10/MRE/09/28).

Participants performed multiple repetitions of five exercises: walking (WLK), jogging (JOG), double- and single-leg squatting (DLS & SLS), and single-leg hopping (HOP). Quiet standing (QuSt) for at least 5 seconds was also recorded to provide reference data for parameters used in this study. Standardised instructions were given on how to carry out the activities. For WLK and JOG participants were asked to walk or jog along a 15 metre walkway at their self-selected 'normal' speed. Steps onto a force platform in the middle of the walkway were used for the analysis to ensure that acceleration and deceleration at the start and end of the trials were excluded. For DLS and SLS participants were instructed to repeatedly squat to their self-determined maximum depth and then return to their starting position. For HOP individuals were asked to hop their self-determined maximum distance and regain their position of quiet single leg standing after landing. Participants were allowed to practice to check that instructions were well understood and executed. For squatting and hopping participants were free to position and move their arms as normally required. We find that patients often need arm movement as a means to maintain control and therefore we measured arm movement in these healthy control subjects rather than imposing a standardised arm position. Participants were asked to perform eight DLS, SLS and HOP trials and five WLK and JOG trials; the first four trials successfully recorded for each activity were analysed. Individuals were given time to rest between trials. Task performance was quantified for each task as this could influence movement strategies and knee control. Velocity (m/s) was determined for WLK and JOG; squat depth by means of CoM vertical displacement (m) for DLS and SLS; and hopped distance by means of IL heel marker horizontal displacement (m) for HOP.

Kinematic data were collected at 250 Hz using an eight camera VICON MX3+ motion analysis system (Oxford Metrics Group Ltd., Oxford., UK). Reflective markers were placed using the VICON PlugInGait full body marker set. Two additional markers were placed on the left and right lateral sides of the iliac crest (LILC and RILC). A static anatomical calibration trial was collected on each participant. The knee axes were aligned using the anatomical calibration trial. In some trials the trunk and hips flexed as such that the markers on the left and right anterior superior iliac crests (LASI and RASI) were occluded; these gaps were filled using the data of the LILC and RILC markers in a custom written programme in Vicon BodyBuilder for Biomechanics (version 1.2, Oxford Metrics Group Ltd., Oxford, UK). Kinematic calculations were performed within VICON Nexus software (version 1.6.1, Oxford Metrics Group Ltd., Oxford., UK) and data were further processed and analysed in Matlab R2012b (The Mathworks Inc., Natick, USA). Marker data were filtered with a fourth order Butterworth filter and a low pass cut off frequency of 20 Hz. Landmarks were used to represent key kinematic components of the body: C7 (for upper trunk); wrist joint centres (arms); pelvis centre; knee joint centres; and ankle joint centres (Figure 1). We explored the head as another landmark but did not include this in the final analysis because the results were very similar to the C7 landmark. Using VICON PlugInGait the CoM was calculated as the weighted average of all body segments. The arm and leg on the weight-bearing side were labelled as the ipsilateral (IL) side and the other arm and leg as the contralateral (CL) side. These labels were kept the same for DLS. Squat initiation and termination were determined automatically as the point at which the CoM moved beyond 95% of its height measured during quiet standing. Bespoke Matlab procedures were written for this calculation.

Time periods used for the kinematic analysis in each task were between IL foot strike and foot off for WLK and JOG; between squat initiation and squat termination for DLS and SLS; and between IL foot strike and peak knee flexion for HOP. The output variables calculated to assess motor control of each movement component were fluency and displacement variability. Fluency was calculated as the inverse of the average number of times per second a landmark velocity in the frontal plane crossed zero (Figure 2). The inverse (Period (s): T=1/f) was used so that a larger value agreed with a more fluent movement [13]. Displacement variability was defined as the standard deviation of the frontal plane trajectory. Frequently, the CoM was graphically observed to predominantly move in one direction during a trial, either medial or lateral. In terms of the movement trajectories this means that the slope was either negative or positive (for instance see Figure 3: WLK trajectory at the bottom of the cluster of four walk trials pointed out by the arrow). The magnitude of this slope was used to extend the analysis of CoM displacement variability. For each trial of each task a line was fitted using regression through the time series of the CoM frontal plane movement to identify a consistent movement trend. Absolute slope values were used for further analysis to avoid the effect of cancelling out of different directions. A value close to zero would indicate that the overall slope was negligible. Variability resulting from each slope was calculated as the standard deviation of the points on the fitted line.

The results were expressed as a percentage of CoM displacement variability and were reviewed descriptively to support interpretation.

For descriptive statistics means and standard deviations are presented. After checking for normal distributions, statistical differences for the main variables (fluency and displacement variability) were analysed. A repeated measures ANOVA was used to investigate CoM differences between WLK and the other tasks. An alpha level of $p<0.05$ was used to evaluate significant overall between-task differences and a Bonferroni adjusted alpha level of $p<0.0125$ was used to explore JOG, DLS, SLS, and HOP compared to WLK. Subsequently, for each individual movement task the eight body landmarks were compared to CoM. An alpha level with Bonferroni adjustment of $p<0.005$ was used. Landmark of most interest was the weight-bearing knee joint centre (Knee IL).

Results

Table 1 lists the average task performance of the five activities quantified by movement velocity and distance; as appropriate. Figure 3 shows the time series of 4 trials of all studied activities, including quiet standing, for a typical subject. As expected, the length of the trajectories representing task duration were uneven (mean ± standard deviation: WLK: 615.2 ± 28.3 ms; JOG: 270.6 ± 32.9 ms; DLS 2,164.3 ± 731.4 ms; SLS: 1,958.4 ± 526.7 ms; HOP: 1,540.2 ± 583.8 ms; QuSt: 910.2 ± 244.3 ms). The graph also illustrates that the subject varied quite substantially between trials in the execution of the same task.

To provide a reference, mean and standard deviations for fluency and displacement variability were calculated for quiet standing; QuSt CoM fluency was 0.015 ± 0.004 s; QuSt CoM displacement variability was 0.394 ± 0.383 mm; very similar results were observed for the other studied landmarks. Compared to quiet standing fluency and displacement variability values were much larger for all dynamic tasks, although these differences were not tested using inferential statistics. More importantly, the overall test using repeated measures ANOVA showed that the difference between the five functional movements was highly significant for CoM fluency ($p<0.001$) and CoM displacement variability ($p<0.001$). Furthermore, compared to WLK all the other four tasks were significantly different for fluency ($p<0.0125$); for displacement variability only JOG was significantly different from WLK ($p<0.0125$) but no significant differences were found for DLS ($p=0.266$), SLS ($p=0.03$) or HOP ($p=0.026$). Figure 4 shows that WLK demonstrated the highest CoM fluency and DLS the lowest. JOG CoM displacement variability was the lowest and although HOP CoM displacement variability was the highest this was not significantly different from WLK.

Table 2 lists fluency results for all landmarks used in the analysis. If the difference compared to CoM was significant ($p<0.005$) this is indicated by the presence of an arrow; the arrow direction is used to illustrate the direction of the significant difference. The hypothesis that fluency would be lower for the constituent components compared to the CoM was explored. In the table the results meeting this expectation are indicated by a down-facing arrow. Only for WLK was the hypothesis more or less upheld that CoM fluency was statistically larger than fluency of the constituent parts; except for C7. Movement at C7 demonstrated significantly larger fluency than the CoM and therefore changed direction less frequently. Ankle, knee and pelvis landmarks all showed less fluency than CoM. For the rest of the activities no pattern emerged consistent with predictions. In the large majority of cases fluency of the landmarks was higher than CoM fluency. Only the IL Ankle (SLS; HOP), IL Knee (JOG; HOP) and Pelvis (HOP) sometimes showed significantly less fluency than CoM (Table 2). IL Knee fluency was significantly lower than CoM for WLK ($p<0.001$), JOG ($p<0.005$), and HOP ($p<0.001$); significantly higher for DLS ($p<0.001$); and not significantly different for SLS ($p=0.05$).

Table 3 lists the displacement variability results. The hypothesis that displacement variability would be larger for the constituent components compared to the CoM was explored. Again, significant differences are indicated by an arrow ($p<0.005$) also illustrating the direction of a significant difference. The results meeting our prediction are indicated by an up-facing arrow. In the large majority of cases, the hypothesis was confirmed. Although in three cases, IL Knee (WLK, $p=0.068$; JOG, $p=0.213$) and Pelvis (SLS, $p=0.142$), differences were in the predicted direction these were not significant and therefore not considered (Table 3). The IL Ankle did not conform to this overall pattern of results, for any of the activities; this also applied to the CL Ankle for DLS when both feet were weight-bearing. Arm (Wrist) displacement variability was consistently high if not the highest. For HOP CL Ankle displacement variability was also substantial; especially when compared to CoM values.

Figure 3 illustrates that movement direction for activities such as WLK or JOG can show a consistent slope in the medial or lateral direction. The effect this might have on our displacement variability results was therefore explored descriptively. For each trial a medial or lateral movement slope was determined with linear regression. The average slope angle approached zero for quiet standing (0.4 ± 0.2 mm/s; mean ± standard deviation). Results were similar for DLS (0.2 ± 0.9 mm/s); SLS (0.5 ± 1.7 mm/s); and HOP (0.9 ± 16.2 mm/s); HOP demonstrated a large standard deviation for this parameter. The average slopes for WLK (18.2 ± 23.1 mm/s) and JOG (45.3 ± 37.3 mm/s) were largest and standard deviations seemed large particularly for JOG. These results were not tested for significant differences, but the influence a slope can have on displacement variability calculations was explored. The variability calculated for the slopes amounted to 70% of overall WLK displacement variability. This same calculated resulted in 91% for JOG; 30% for DLS; 38% for SLS; and 74% for HOP. Therefore the influence of observed movement trends in the medial or lateral direction on displacement variability results seems to be relevant to consider in the interpretation of these movements.

Discussion

In this study we explored fontal plane displacement variability and fluency during five functional movements and used a simplified eight component kinematic model of the whole body to explore coordination in relation to CoM control. During all movement tasks, CoM displacement variability was smallest compared to the constituent components. Compensatory variability through inter-limb movement coordination must have occurred to compensate for each other with the effect that CoM displacement variability in the frontal plane was well controlled. There were significant differences between tasks for CoM displacement variability. It appears that whilst the movement solution might have differed between tasks, the overall rule of moving with compensatory variability [26] was maintained. For instance, HOP compared to WLK demonstrated large increases in displacement variability of Wrists (271-356%) and CL Ankle (448%) and C7 (305%) whilst the CoM displacement variability was much less increased (130%). In other words it would appear that the organisational principle of compensatory variability is well maintained even if landing from a hop looks more disorganised than walking. Therefore, whole body control in the frontal plane was demonstrated across activities. However, the exact implementation of this strategy varied in terms of fluency and displacement variability. The exception was the weight-bearing ankle with a displacement variability significantly smaller than CoM for WLK, DLS and SLS. Clearly, foot contact with the ground substantially constrained ankle movement.

We did not observe a pattern where CoM fluency was consistently largest compared to the constituent components. In fact, in many cases CoM fluency was significantly smaller. Also, there were many cases where there were no significant differences between fluency of the

CoM and the constituent components. This was not predicted a priori and suggests that the motor control strategy of compensatory variability coincides with fewer changes of movement direction (high fluency) of the constituent components rather than more.

Particularly the IL Knee fluency varied between tasks and was lower than CoM fluency for WLK, JOG and HOP but higher for DLS. There was no significant difference for SLS. Because of this task dependence of fluency and displacement variability each task will briefly be discussed separately to consider the specific implications of the results.

Normal walking involves a steady oscillation of the CoM in the sagittal and frontal plane [29]. This regular change of movement direction with every step will contribute to the calculated fluency. An average WLK CoM fluency of 0.43 is consistent with 2.32 changes of direction per second which on average is the approximate time a stride takes [30]. Therefore predominantly the normal gait cycle oscillation seems strongly represented in this fluency value; the same can be said for CoM displacement variability which was found to be 70% accounted for by the observed overall movement trend during the stance phase of gait. Whilst C7, IL & CL Wrist moved with about the same fluency as CoM during WLK which could indicate synchronised movement with the gait cycle; the displacement variability was significantly larger for these components compared to CoM; consistent with the principle of compensatory variability constraining CoM frontal plane movement and maintaining balance [26; 31]. IL Ankle and IL Knee and Pelvis demonstrated significantly lower fluency which indicates a more frequent change of movement direction compared to the CoM trajectory and could be interpreted as active participation in multiple aspects of motor control: compensatory variability to constrain CoM frontal plane movement and limb support and stability during the stance phase. IL Ankle and Knee displacement variability was lowest compared to the other components. It is reasonable to expect that besides the constraints of foot contact with the floor, a high muscular activation necessary for supporting the body [32] will result in higher stiffness in the weight-bearing limb. This is expected to result in lower fluency and displacement variability for these landmarks. A single task objective for knee movement control does not seem to exist during gait.

The average JOG CoM fluency of 0.16 is much lower than for WLK indicating that the frequency of CoM oscillation is higher during this activity. An average jogging speed of 2.76 m/s relates to approximately 2.67 changes per second due to the jog cycle (average stride time is 0.75s [33]), which would result in a fluency of 0.37. During running compared to gait, step width has been found to be narrowed and the side to side movement of the CoM is reduced [31; 34]. Also, landing on the stance limb will occur without the double support phase seen in gait. Therefore, balance in the frontal plane will be more challenging compared to walking which could explain that CoM fluency is substantially reduced. Furthermore, the larger impact of heel strike may introduce extra oscillations to the CoM trajectory. JOG CoM displacement variability was substantially smaller than for WLK and 91% of this was accounted for by the overall trend of CoM movement. These findings also seem consistent with the narrower CoM trajectory in the frontal plane in JOG compared to WLK.

IL Knee fluency was lowest for JOG compared to the other activities. Furthermore, this was significantly lower than JOG CoM fluency and coincided with relatively low IL Knee displacement variability during JOG. In other words, the knee moved more frequently with smaller displacement variability which could be interpreted to result from an increase in knee stiffness for JOG compared to WLK. It has been demonstrated that higher walking speed coincides with increased knee stiffness [35].

DLS CoM fluency was lowest among the activities under investigation. The absence of a CoM medial or lateral trend demonstrated the movement was on average organised around a midline with the low fluency demonstrating a frequent change of movement direction. The average squat cycle time was 2.16 seconds; this movement pattern alone would have resulted in much higher fluency values than observed. The relatively low CoM displacement variability for this activity indicates that CoM was reasonably constrained. The double leg support clearly offered good overall balance for this activity. IL Knee fluency was largest for this activity with the largest IL Knee displacement variability. The knee movement trajectories showed that knees were moved outward during flexion and moved back inward during extension (Figure 5). Therefore the overall movement pattern will account in large part for the large knee displacement variability observed. This relatively large overall knee movement would hide minor oscillations in the frontal plane; alternatively, if balance was not particularly challenged during this activity knee movement could be less involved with the overall balance.

SLS obviously is a more challenging task compared to DLS [14]; on average CoM squat depth was 0.1m (27%) less for SLS than for DLS. The average squat cycle time was 1.96 seconds; very similar to DLS. SLS CoM fluency was significantly reduced compared to WLK with values close to JOG and DLS. SLS CoM displacement variability was relatively large which was not unexpected. This can be considered consistent with a reduced frontal plane stability during the exercise certainly compared to DLS. Displacement variability of C7, CL Ankle and Knee and Wrists was relatively large which may relate to their substantial involvement in the movement task and balance control within the context of compensatory variability. IL Knee fluency was not significantly different from CoM fluency, but IL Knee displacement variability was significantly larger than CoM displacement variability. The demands of the movement task to provide balance control in concert with the rest of the body seem to have become more dominant compared to the requirement to provide a substantial support moment through limb stiffness.

HOP could be considered the most challenging task in this study [14]. CoM fluency was significantly smaller than WLK but about twice the values found for JOG, DLS, and SLS. HOP CoM displacement variability was the largest compared to all other activities. An increasing challenge for overall balance seems to result in a larger CoM displacement variability with a reduced frequency of movement changes (higher fluency). HOP IL Knee fluency was significantly lower and IL Knee displacement variability was significantly increased compared to CoM. This would be difficult to explain only by the mechanism of stiffness regulation. The requirement for overall balance control clearly places higher demands on the IL Knee.

Generally speaking, the task influenced the movement solution of the complex motor system which impacted on fluency and displacement variability of the individual components. For the purposes of our study into the effect of knee injury on performance of functional movement it is of interest to specifically focus on the weight-bearing knee. The role of the IL Knee in the studied movement tasks seemed to be multiple. Obviously the knee will be involved in creation of the movement itself and will require to be kept stable during dynamic activities. In addition, providing the support moment during weight-bearing and contributing to the overall balance and efficiency of total body movement are important roles. It can be argued that fluency is related to movement frequency (smoothness), while displacement variability is related to movement amplitude. The results for fluency tended to be less consistent and more difficult to interpret; e.g., IL Knee fluency was significantly lower than CoM for WLK, JOG, and HOP; significantly higher for DLS; and higher but not significantly for SLS. Each task places its own demands on knee movement control which seems to alter fluency based on which role of knee movement is most important. The results for displacement variability were more consistent and the parameter fits better with the motor control literature [19], particularly the concept of compensatory variability.

Fluency and displacement variability together can be used to give an idea how knee movement is controlled and therefore how patients may adapt their movement strategies. The pattern of changes in IL Knee fluency and displacement variability between activities did not mimic the CoM changes. This may relate to the relative importance of the various roles the knee has for each movement task and the challenges to knee stability and overall balance resulting from the task. Clearly a single knee control strategy was not identified and therefore movement adaptations observed in patients will need to be explored and explained in their own context for each functional exercise used in rehabilitation. Besides task constraints of the overall movement [36], control of knee movement presumably has its own constraints particularly with respect to weight-bearing and joint stability. Knee fluency and displacement variability therefore need to be interpreted within the context of the constraints most relevant for this joint for a specific task. This may go some way to explain the seemingly conflicting results we found so far in knee fluency in knee patients [13-14]. The next step in this research will be to apply this framework of interpretation in a comparison of ACLD and ACLR patients to this group of healthy subjects. Further research could also explore in-depth the inter-individual differences in the movement solutions used for such rehabilitation exercises.

Conclusions

Five functional exercises used in rehabilitation were studied kinematically to explore whole body coordination and knee movement control in the frontal plane. Fluency of movement and displacement variability were used in this analysis. Evidence of a whole body strategy was demonstrated where CoM displacement variability was smallest compared to the constituent landmarks. The results for fluency were not consistently the same across the movement tasks. Task dependence was particularly demonstrated for the weight-bearing knee in terms of its movement fluency and displacement variability. This may relate to the relative importance of the various roles the knee has for each movement task specifically the challenges to knee stability and weight-bearing as well as the overall movement and balance requirements of the task.

List of Abbreviations

WLK: walking task; JOG: jogging task, DLS: double-leg squatting task; SLS: single-leg squatting task HOP: hopping task; QuSt: Quiet standing; LILC and RILC: left and right lateral sides of the iliac crest; LASI and RASI: left and right anterior superior iliac crests; C7: 7th cervical spinous process; CoM: Centre of Mass; IL: ipsilateral; CL: contralateral.

Competing interests

The authors declare that they have no competing interests.

Author's contributions

All authors have made substantial contributions to the conception and design of this research and this manuscript and have provided substantial intellectual input. PR and KB have been involved in data acquisition and processing. KB has provided substantial support with subject recruitment. RvD has performed all data analysis and prepared figures and tables for this paper. RvD has drafted the manuscript. All authors have been involved in revising the manuscript critically and have given final approval of the version to be published.

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Figure Legends

Figure 1

Kinematic model used in the analysis of fluency and variability. The eight landmarks on the body are considered part of the complex system that controls the CoM in the frontal plane.

Figure 2

Example of a time series of frontal plane movement used to determine fluency (based on velocity) and variability (based on position). The dots on the velocity plot indicate the instances when the velocity crossed the zero line.

Figure 3

Time series of the CoM position in the frontal plane for a typical subject for the four trials used in the analysis. The time phase of the movement (stance, landing, weight-bearing) over which the analysis was applied is displayed. All positions were calculated relative to the ipsilateral (right) ankle joint centre (straight, solid line) at footstrike. During double support tasks the contralateral (left) ankle position was in the region of 0.3-0.4m; therefore off the figure. The zero point in time represents foot strike for WLK (thin, dashed line), JOG (thick, solid line) and HOP (thick, dash-dotted line) and start of the task for DLS (thick, dashed line) and SLS (thin, dotted line).

Figure 4

Fluency and variability results for CoM and ILKnee between the five activities used in the analysis. Means and standard deviations are displayed. * indicates whether ILKnee results were significantly different from CoM results (* $p<0.05$).

Figure 5

Time series of the position in the frontal plane of the CoM and knee joint centres during DLS for the same typical subject (Figure 3) for the four trials used in the analysis. Note that both knees move outwards in the first half of the movement and back inwards in the second half of the movement.

Figure 1

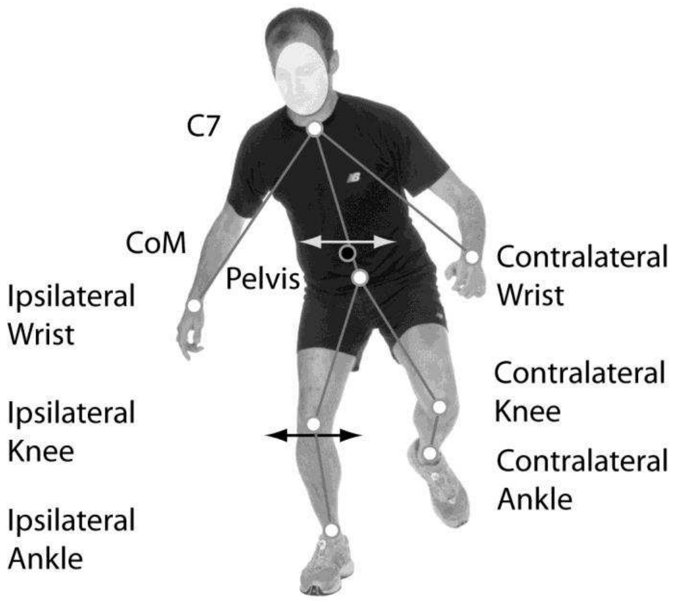

C7
CoM
Pelvis
Ipsilateral
Wrist
Contralateral
Wrist
Contralateral
Knee
Ipsilateral
Knee
Contralateral
Ankle
Ipsilateral
Ankle

Figure 2

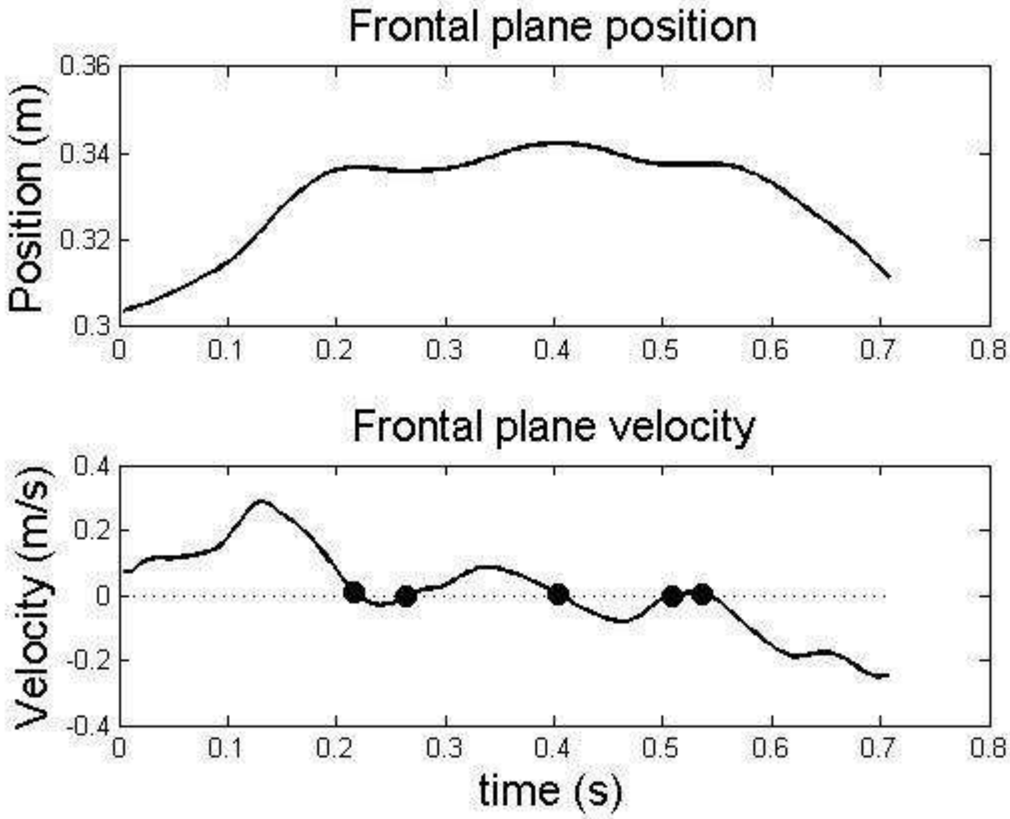
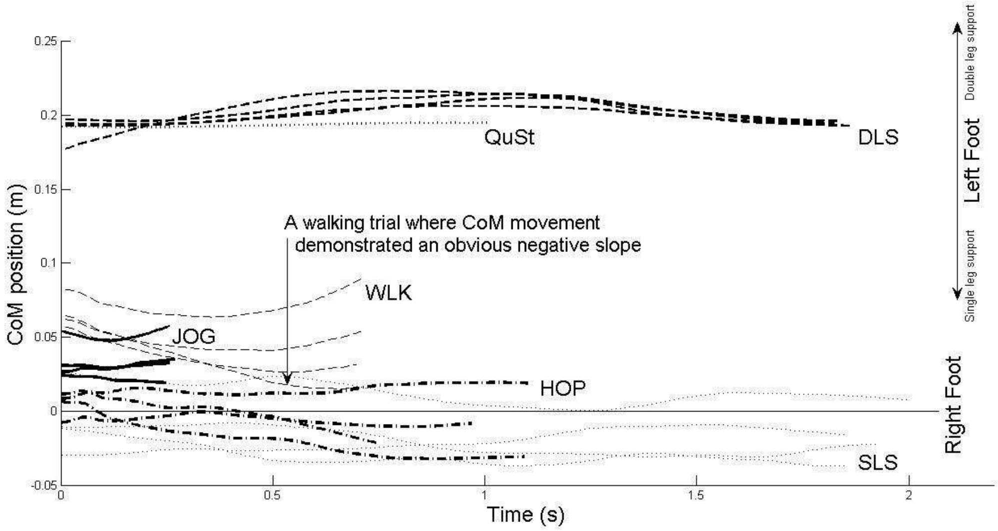

Frontal plane position
Position (m)
0.36
0.34
0.32
0.3
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
0.8
Frontal plane velocity
Velocity (m/s)
0.4
0.2
0
-0.2
-0.4
time (s)

Figure 3



0.25
0.2
0.15
0.1
0.05
0
-0.05
CoM position (m)
0
0.5
1
1.5
2
Time (s)
QuSt
DLS
A walking trial where CoM movement demonstrated an obvious negative slope
WLK
JOG
HOP
SLS
Double leg support
Left Foot
Single leg support
Right Foot

Figure 4

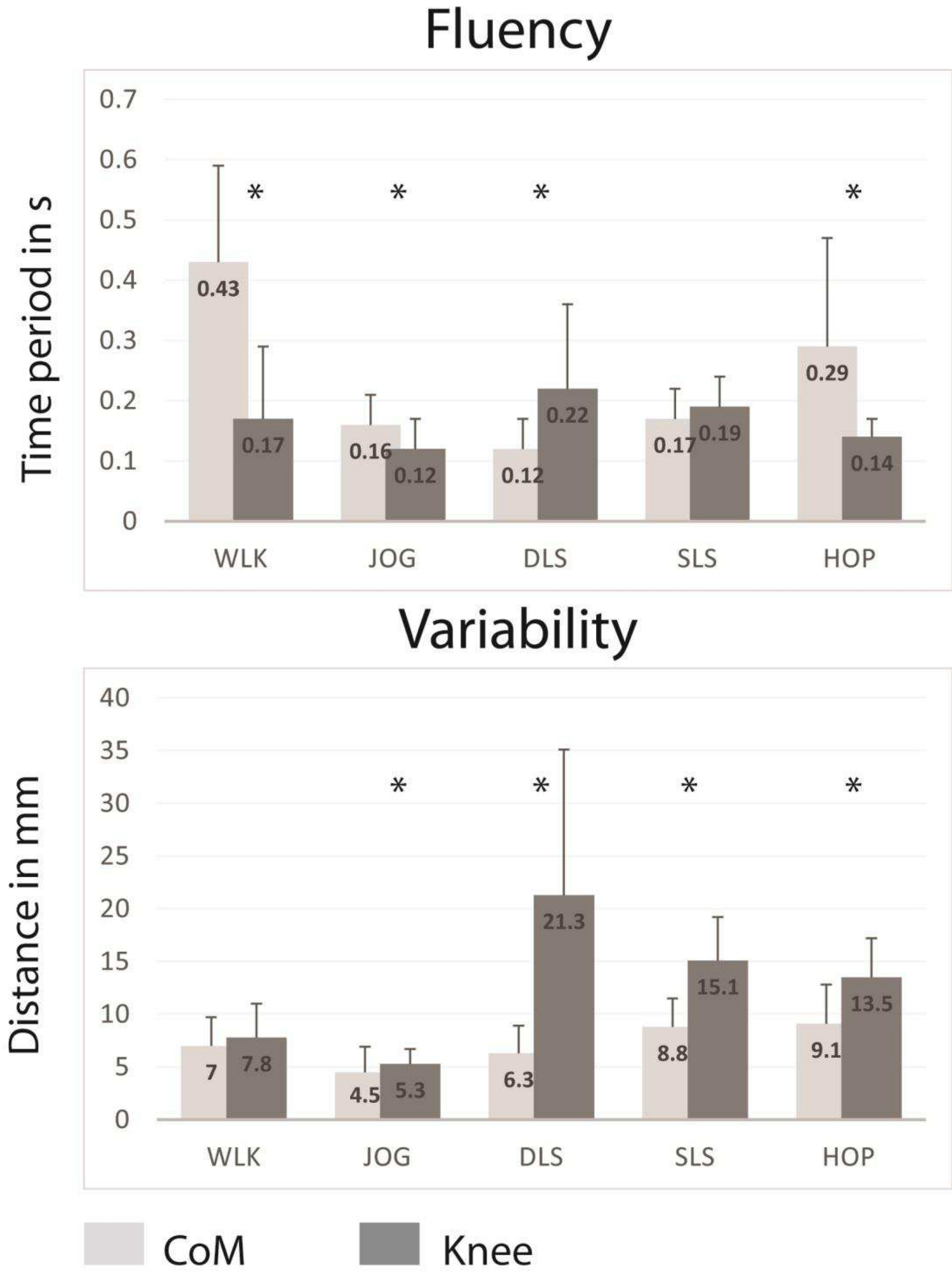

Fluency
Time period in s
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
*
*
*
*
0.43
0.17
0.16
0.12
0.12
0.22
0.17
0.19
0.29
0.14
WLK
JOG
DLS
SLS
HOP
Variability
Distance in mm
40
35
30
25
20
15
10
5
0
*
*
*
*
7
7.8
4.5
5.3
6.3
21.3
8.8
15.1
9.1
13.5
WLK
JOG
DLS
SLS
HOP
CoM
Knee

Figure 5

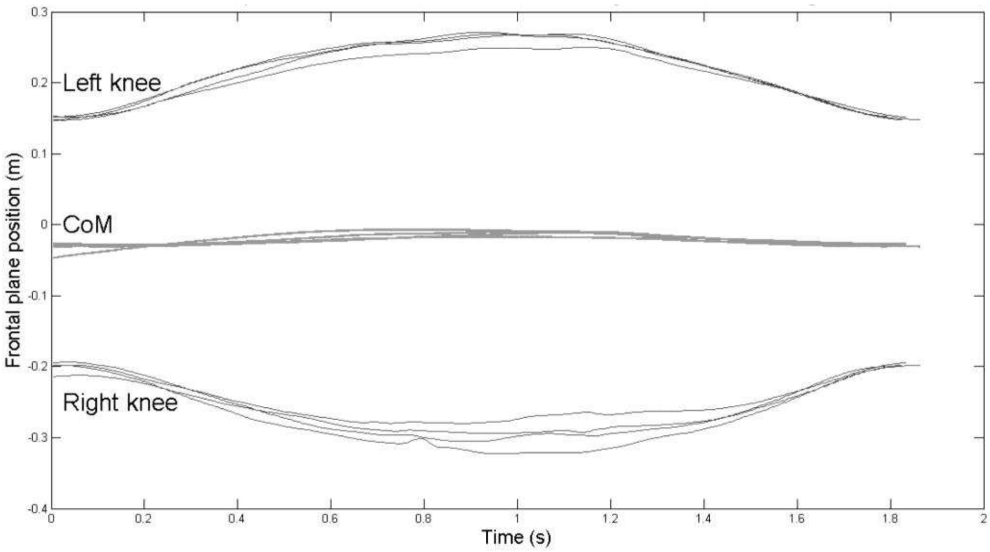

Left knee
CoM
Right knee
Frontal plane position (m)
Time (s)
0.3
0.2
0.1
0
-0.1
-0.2
-0.3
-0.4
0
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8
2

Table 1: Demographic information and task performance parameters for the participating group of healthy subjects. Means and standard deviations are provided.

Parameter	Value
Gender	10F/19M
Age (yrs)	28.0 ± 6.9
Height (m)	1.73 ± 0.11
Mass (kg)	73.5 ± 16.4
WLK velocity (m/s)	1.45 ± 0.17
JOG velocity (m/s)	2.76 ± 0.45
DLS squat depth (m)	0.37 ± 0.10
SLS squat depth (m)	0.27 ± 0.08
HOP distance (m)	1.36 ± 0.31

Table 2 Fluency of all landmarks used in this analysis. Mean and standard deviations are shown. Differences between tasks for CoM were all highly significant ($p<0.001$). Significant comparisons between landmarks and CoM are indicated by arrows: ↓= value significantly lower than CoM; ↑= value significantly higher than CoM ($p<0.005$).

Fluency (s)	**WLK**	**JOG**	**DLS**	**SLS**	**HOP**
CoM	0.43±0.16	0.16±0.05	0.12±0.05	0.17±0.05	0.29±0.18
Ankle IL	0.14±0.06 ↓	0.17±0.06	0.12±0.06	0.13±0.04 ↓	0.11±0.02 ↓
Knee IL	0.17±0.12 ↓	0.12±0.05 ↓	0.22±0.14 ↑	0.19±0.05	0.14±0.03 ↓
Pelvis	0.25±0.13 ↓	0.14±0.07	0.21±0.09 ↑	0.27±0.08 ↑	0.15±0.04 ↓
C7	0.57±0.12 ↑	0.25±0.04 ↑	0.22±0.13 ↑	0.38±0.13 ↑	0.44±0.13 ↑
Ankle CL	0.20±0.10 ↓	0.23±0.06 ↑	0.13±0.05	0.32±0.12 ↑	0.36±0.11
Knee CL	0.16±0.07 ↓	0.19±0.07	0.23±0.09 ↑	0.29±0.08 ↑	0.31±0.07
Wrist IL	0.40±0.14	0.27±0.03 ↑	0.22±0.10 ↑	0.37±0.15 ↑	0.38±0.09
Wrist CL	0.41±0.14	0.25±0.07 ↑	0.24±0.14 ↑	0.32±0.16 ↑	0.36±0.08

Table 3 Variability of all landmarks used in this analysis. Mean and standard deviations are shown. Differences between tasks for CoM were all highly significant ($p<0.001$). Significant comparisons between landmarks and CoM are indicated by arrows: ↓= value significantly lower than CoM; ↑= value significantly higher than CoM ($p<0.005$).

Variability (mm)	**WLK**	**JOG**	**DLS**	**SLS**	**HOP**
CoM	7.0±2.7	4.5±2.4	6.3±2.6	8.8±2.7	9.1±3.7
Ankle IL	5.5±1.4 ↓	7.3±3.0 ↑	3.4±2.6 ↓	4.9±3.5 ↓	8.0±4.0
Knee IL	7.8±3.2	5.3±1.4	21.3±13.8↑	15.1±4.1 ↑	13.5±3.7 ↑
Pelvis	11.6±3.3 ↑	6.4±2.3 ↑	8.8±3.0 ↑	9.7±4.2	16.2±4.9 ↑
C7	12.1±3.0 ↑	8.8±2.6 ↑	8.9±3.2 ↑	24.4±11.2↑	36.9±12.5↑
Ankle CL	14.9±6.8 ↑	11.8±4.1 ↑	3.6±2.8 ↓	34.5±22.5↑	66.8±33.6↑
Knee CL	15.4±7.7 ↑	10.7±5.0 ↑	24.2±15.4↑	24.3±14.1↑	46.2±17.5↑
Wrist IL	24.2±13.9↑	34.8±9.7 ↑	20.8±14.5↑	36.6±22.1↑	86.1±34.4↑
Wrist CL	29.6±19.2↑	36.0±15.9↑	22.7±17.3↑	36.7±22.4↑	80.2±33.3↑