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# Detecting and quantifying global instability during a dynamic task using kinetic and kinematic gait parameters

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ABSTRACT

*Objectives:* Instability during gait can be identified in many different ways. Recent studies have suggested utilizing spatiotemporal parameters to detect instability during gait. Detecting instability using kinetic and kinematic gait parameters has not yet been examined fully. In addition, these studies have not yet identified measures that are capable of assessing the magnitude of instability. The objective of the present study was to identify kinetic and kinematic gait parameters that can best identify instability and quantify its magnitude.

*Methods:* Ten healthy men underwent successive gait analysis testing under three controlled settings: (1) Stage 0 instability (control setting), (2) Stage 1 instability and (3) Stage 2 instability. The levels of instability were precisely applied with the use of a controlled perturbation device (AposTherapy System). Differences between all stages and between stages were identified using Friedman and Wilcoxon tests.

*Results:* Stride-to-stride variability (STSV) in kinetic and kinematic measures increased significantly between stages 0 and 1 or between stages 0 and 2 for almost all parameters (all $P<0.05$). A significant increase between stage 0 and both stages 1 and 2 was found for knee flexion moment, knee varus moment, knee flexion angle and hip adduction angle. The increase between stages 1 and 2 was variable. Only the knee varus moment parameter showed a significant increase in STSV between stages 1 and 2 ($P=0.026$).

*Conclusions:* Almost all kinetic and kinematic gait parameters are sensitive to changes in global instability in a dynamic task. The most sensitive are parameters measured at the knee. Of these, STSV in knee varus moment can be used to quantify the magnitude of dynamic instability.



## 1. Introduction

Instability during gait has received considerable focus over the last several years due to its association with falling. Falling is a common and dangerous problem in society. In the elderly, falling is particularly prevalent and can be incapacitating when it occurs (Voermans et al., 2007). Falling is only predicted to increase in frequency as life expectancy continues to rise (Brauer et al., 2000). Instability and falling is also prevalent in patients with osteoarthritis (Arnold and Faulkner, 2007), Parkinson's disease (Bloem et al., 2001; Factor et al., 2011; Plotnik et al., 2011), Huntington's disease (Grimbergen et al., 2008), cerebral palsy (Tsirikos et al., 2003) and other neurological disorders.

Instability is classically defined according to the relationship between a person's center of mass (COM) and base of support (BOS) (Winter, 1995). The further the COM is from the BOS, the more "unstable" the person (Winter, 1995). During gait, however, the situation becomes much more complex. During each gait cycle the location of the COM of the body follows a sinusoidal curve between both feet. The curve usually fits within the dynamic BOS created by footsteps during gait (Winter, 1995). In subjects who are unstable in gait, the COM curve creeps beyond the BOS defined by the feet until it reaches a maximum at which the patient is at risk for falling (Winter, 1995). Due to its complex manifestation in gait, stability in gait is often separated into local and global classifications (Dingwell et al., 2000). Local dynamic stability refers to the body's ability to recover from small perturbations. It can be quantified using Lyapunov exponents (LyE) (Arellano et al., 2009). On the other hand, global dynamic stability refers to the body's ability to recover from large-scale perturbations, such as slip or trip (Dingwell and Cavanagh, 2001).

Several techniques have been developed in order to identify global instability before a fall occurs. Many of these include static tests such as quiet standing and retropulsion tests (Bloem et al., 1998). Over the

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last decade, however, research has shown that dynamic instability is vastly different from static instability (Brauer et al., 2000). In addition, most falls occur during dynamic motion (i.e. simple walking) (Voermans et al., 2007). For this reason, researchers have attempted to find measures of dynamic instability. Some researchers have measured instability by comparing the COM to the BOS during gait (Lee and Chou, 2006). In one of the first works in the field, Guimaraes and Isaacs showed that subjects who are unstable constantly adopt different walking patterns and that gait variability can be an accurate tool in identifying instability (Guimaraes and Isaacs, 1980). Since then a number of studies have examined the stride-to-stride variability (STSV) in fallers compared to non-fallers. Studies on the elderly have shown that STSV increases with age and frequency of falling (Grabiner et al., 2001; Hausdorff et al., 2001a,b; Hollman et al., 2007a,b). This has also been confirmed in patients with basal ganglia disorders (Hausdorff et al., 2003; Hausdorff et al., 1998; Schaafsma et al., 2003). All these studies have examined STSV in spatiotemporal parameters of gait (e.g. step length, velocity and single-limb-support). Joint kinetic and kinematic parameters of gait that can identify instability have not yet been examined. In addition, these studies have not been able to quantify the severity of dynamic instability using these gait parameters. Instead, they have used STSV as only a marker of instability (Hausdorff et al., 2001a,b) and as markers for increased stability after therapy (Hausdorff et al., 2001a,b).

The present study was therefore devised to identify kinetic and kinematic gait parameters that can identify global instability in a dynamic task, as well as determine which parameters, if any, can quantify the severity of instability as it is increased in a controlled, step-wise fashion in otherwise healthy, stable individuals. The dynamic properties of STSV can be examined by looking at the difference in the timing of events between trials or in the consistency in the values of gait parameters from stride-to-stride (Hausdorff et al., 2001a,b). We chose to examine the consistency in the values from event to event (average and standard deviation of the peaks for each parameter) of kinetic and kinematic parameters themselves rather than the timing between events.

## 2. Methods

### 2.1. Participants

The study cohort was comprised of 10 healthy male undergraduate students with equivalent shoe size (French 43). Participants had an age of $25.0 \pm 2.1$ years, height of $178.1 \pm 3.4$ cm and weight of $74.4 \pm 3.9$ kg. Exclusion criteria were any orthopedic, musculoskeletal or neurological pathology. Approval of the institutional review board was obtained and all participants gave informed consent prior to entering the study. The purpose and methods of the study were explained to all subjects.

Stages of controlled instability

Three stages of instability were assessed in the present study. Each stage was precisely applied with the use of a controlled perturbation device (AposTherapy System). The device consists of a foot-worn platform to which two convex shaped biomechanical elements, constructed from shoe sole material (Appendix A in supplementary material), can be attached (Fig. 1). The convex nature of the elements induces instability during walking. The convexity of the elements increases as the height of the elements increases. Therefore, if the height of the elements is increased, the instability is increased as well. A variety of elements of different heights can be attached and interchanged on the device. Stage 0 of instability (control setting) was defined as the device without any attached elements. Effectively this was assumed to be similar in structure to a regular walking shoe. Stage 1 of instability was defined as the device with elements with a height of 9.2 mm. Stage 2 of instability was defined as the device with elements with a height of 10.8 mm.

### 2.2. Experimental protocol

Each subject was fitted with the device at each stage of instability by a trained physiotherapist. Successive gait analysis testing was performed at each stage.

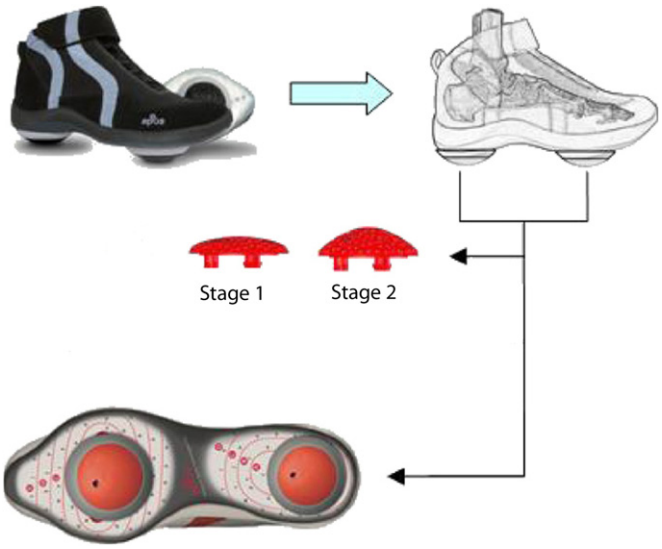


**Fig. 1.** The biomechanical system used to apply the controlled stages of instability. The biomechanical device (AposTherapy System) is a foot-worn platform with a specially designed sole that is capable of attaching to two rubber convex elements. The elements generate perturbation and instability when the device is worn. The elements come in different heights. For the purposes of the present study, Stage 0 of instability was defined as the platform without any elements (not shown), Stage 1 of instability was defined as the platform with elements of size 9.2 mm attached at the sole and Stage 2 of instability was defined as the platform with elements of size 10.8 mm attached at the sole.

The subjects were asked to walk with the apparatus with elements at a self-selected velocity, which was indicated by a metronome to ensure consistent cadence throughout the trials. The metronome was set specifically to each subject before data acquisition. The study controlled for cadence in order to minimize the effects of its variation on the STSV in other gait parameters. Without controlling for the normal variation in cadence, the STSV in the kinetic and kinematic parameters would be much more difficult to discern. Six trials at each stage were collected per subject for STSV. All conditions were tested in a random order on the same day.

### 2.3. Data acquisition, processing and analysis

Gait analysis of each subject was performed as described in previous studies (Haim et al., 2012; Haim et al., 2011) (Appendix B in supplementary material). A Vicon motion analysis system (Oxford Metrics Ltd., Oxford UK) accompanied by two three-dimensional AMTI force plates was used for data capture. A standard reflective marker set was used to define joint centers and axes of rotation (Kadaba et al., 1990). The dominant leg was chosen for all patients for consistency and to control for the inherent differences between limbs.

The following parameters were measured: Knee Flexion Moment, Knee Varus Moment, Ankle Dorsiflexion Moment, Ankle Inversion Moment, Hip Extension Moment, Hip Adduction Moment, Knee Flexion Angle, Knee Extension Angle, Knee Varus Angle, Ankle Dorsiflexion Angle, Ankle Inversion Angle, Hip Extension Angle, and Hip Adduction Angle. The data were graphed for the stance phase of each trial for every stage of instability. In order to obtain a measure of global instability, the average and standard deviation of peaks of the graphs for all parameters were calculated.

All variables were tested by the Kolmogorov–Smirnov test for normal distribution. The STSV was defined as the standard deviation of the peaks at each stage of instability. The STSV was compared between stages using the Wilcoxon Rank nonparametric test. As a secondary objective of the study, the averages of the peaks were also compared between stages to determine if any noteworthy trends are evident as instability increases. The averages were compared across stages using the Friedman nonparametric test. All statistical tests were carried out in SPSS v.17 by a biostatistician.

## 3. Results

The patients' self-selected velocities ranged from $1.15 \pm 0.16$ m/s. The STSV was lowest at stage 0 of instability (control setting) for all gait parameters aside from ankle inversion moment. There was a significant increase in STSV with instability in all parameters aside from ankle inversion moment and ankle inversion angle (Table 1).

**Table 1**
Changes in stride-to-stride variability (STSV) across three stages of instability.

| Parameter | Stages of instability | | | Significance | | |
|---|---|---|---|---|---|---|
| | 0 | 1 | 2 | 1 > 0 | 2 > 0 | 2 > 1 |
| Knee flexion moment | 60.0 | 120.8 | 125.4 | 0.0065* | 0.004* | 0.4765 |
| Knee varus moment | 52.5 | 77.4 | 89.2 | 0.011* | 0.004* | 0.0255* |
| Ankle dorsiflexion moment | 50.3 | 78.6 | 63.1 | 0.057 | 0.033* | 0.998 |
| Ankle inversion moment | 52.6 | 39.2 | 35.4 | 0.998 | 0.998 | 0.998 |
| Hip extension moment | 67.6 | 93.9 | 86.1 | 0.193 | 0.033* | 0.998 |
| Hip adduction moment | 43.8 | 67.4 | 66.3 | 0.085 | 0.0055* | 0.998 |
| Knee varus angle | 0.64 | 2.2 | 2.1 | 0.0465* | 0.0105* | 0.998 |
| Knee flexion angle | 1.1 | 1.6 | 2.0 | 0.1205 | 0.0075* | 0.0255* |
| Knee extension angle | 0.86 | 1.4 | 1.7 | 0.1015 | 0.0105* | 0.2205 |
| Hip extension angle | 0.84 | 1.0 | 0.80 | 0.1665 | 0.998 | 0.998 |
| Hip adduction angle | 0.71 | 0.98 | 0.94 | 0.011* | 0.033* | 0.998 |
| Ankle dorsiflexion angle | 0.84 | 1.3 | 0.92 | 0.0465* | 0.297 | 0.998 |
| Ankle inversion angle | 0.61 | 1.1 | 0.89 | 0.057 | 0.157 | 0.998 |

STSV is measured as the standard deviation of the peaks of the kinetic and kinematic parameters; the values of moments are presented in Newton meter/kilogram ( $\times 10^3$); the values of angles are presented in degrees; differences between stages were compared using the Wilcoxon Rank nonparametric test.

* The significance threshold was set at 0.05.

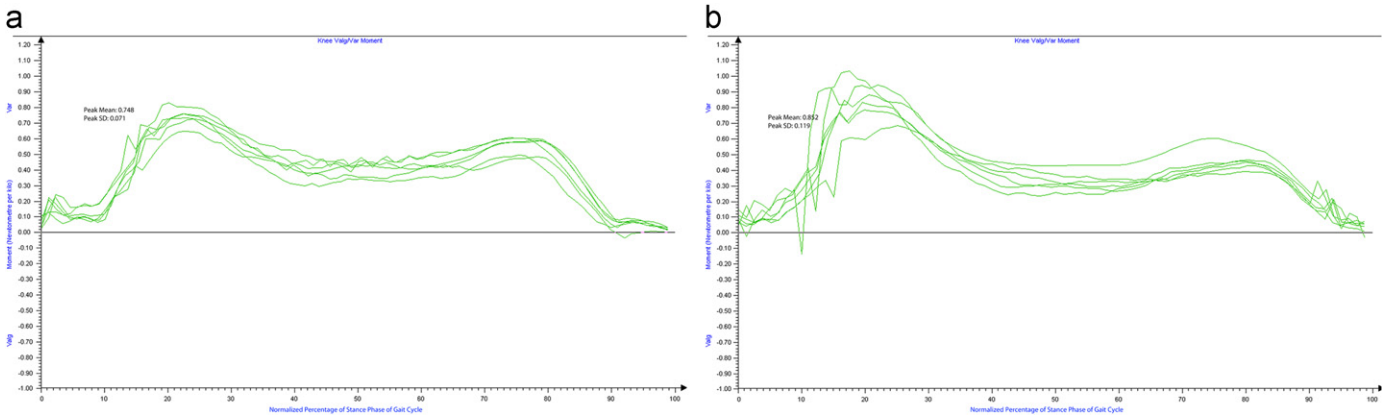


**Fig. 2.** Increase in stride-to-stride variability (STSV) from stages 0 to 2 of instability for knee varus moment in one representative patient. a and b represent the graphs of the knee varus moment of the six trials of a subject for stage 0 and stage 2 of instability, respectively. The study measured the STSV as the standard deviation of the first peak of the graph. As can be seen clearly, the variability (i.e. standard deviation) of the first peak of knee varus moment was significantly greater at stage 2 of instability (b) in comparison to stage 0 of instability (a) ($P=0.026$).

This increase was significant either from stage 0 to stage 1 of instability or from stage 0 to stage 2 of instability. There was a significant increase in STSV from stage 0 to both stage 1 and stage 2 in knee flexion moment, knee varus moment, knee varus angle and hip adduction angle. Fig. 2a&b illustrates the relatively low STSV at stage 0 compared to the higher STSV at stage 2 for knee varus moment from one subject.

The change in STSV form stage 1 to stage 2 of instability was more variable. There was an increase in STSV from stage 1 to stage 2 in knee flexion moment, knee varus moment, knee flexion angle and knee extension angle. This increase, however, was significant for knee varus moment and knee flexion angle ($P=0.0255$). The results of the peak knee flexion moment, knee varus moment and knee flexion angle are illustrated in Fig. 3a–c. Other parameters showed a slight drop in STSV from stage 1 to stage 2. The STSV variable at stage 2, however, was still significantly greater than at stage 0 for all parameters aside from ankle dorsiflexion angle.

There were some noteworthy changes in the averages of the parameters between conditions as well. A significant increase in mean magnitude of the peaks was found across stages of instability in ankle dorsiflexion moment ($P=0.045$), hip adduction angle ($P=0.013$) and ankle inversion angle ($P=0.030$) (Table 2).

## 4. Discussion

The present study aimed to identify the kinetic and kinematic parameters that are able to detect global instability in a dynamic setting. In addition, the study was designed to determine which of these parameters can also quantify the severity of instability. The results of the study found that almost all of the kinetic and kinematic measurements were able to detect instability during gait when a sufficient level is presented. Only ankle inversion moment and ankle inversion angle showed no sensitivity to instability.

Most of the parameters that succeeded at identifying instability found a significant difference in STSV between stage 0 of instability and either stage 1 or stage 2 of instability, but not both. Significant differences between stage 0 and both stages 1 and 2 were observed for only knee flexion moment, knee varus moment, knee varus angle and hip adduction angle. This suggests that these parameters are the most capable at detecting global instability since they are able to identify the presence of instability with precision. Interestingly, however, it appears that many of these parameters may not be sensitive enough at detecting increasing levels of instability since their ability to detect the difference between stages 1 and 2 of instability was much poorer.

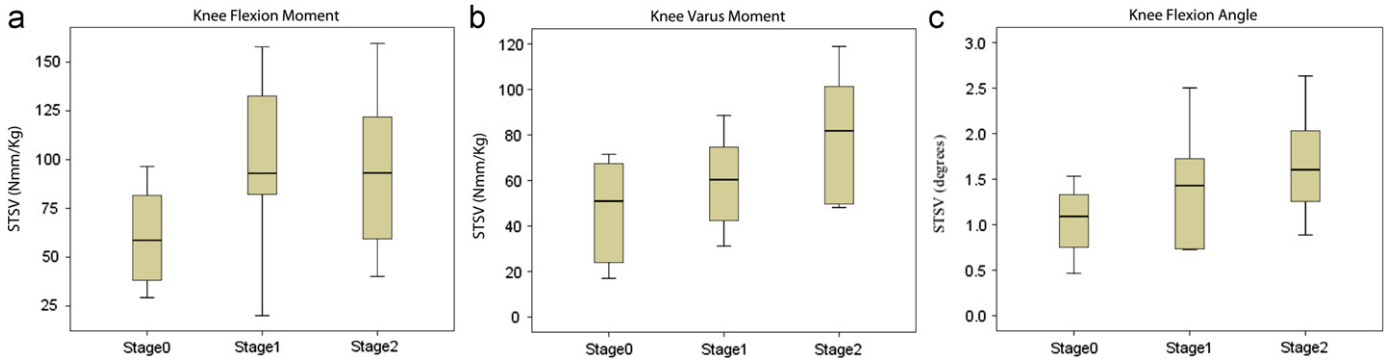


**Fig. 3.** Increase in stride-to-stride variability (STSV) from stage 0 to stage 1 to stage 2 of instability. This figure illustrates the increase in STSV from stage 0 of instability to stage 2 of instability. The difference between stage 0 and stage 1, as well as the difference between stage 0 and stage 2, was significant for all three parameters. Only knee varus moment (b), however, showed a significant increase in STSV from stage 1 to stage 2. This suggests that it is the most sensitive parameter as well as the greatest quantifier of the severity of dynamic instability.

**Table 2**
Changes in mean peak magnitude across three stages of instability.

| Parameter | Stages of instability | | | Significance |
|---|---|---|---|---|
| | 0 | 1 | 2 | *P* Value |
| Knee flexion moment | 0.800 | 0.691 | 0.743 | 0.896 |
| Knee varus moment | 0.565 | 0.760 | 0.683 | 0.121 |
| Ankle dorsiflexion moment | 1.55 | 1.63 | 1.638 | 0.045* |
| Ankle inversion moment | 0.177 | 0.207 | 0.234 | 0.169 |
| Hip extension moment | 1.30 | 1.17 | 1.17 | 0.264 |
| Hip adduction moment | 0.545 | 0.715 | 0.708 | 0.264 |
| Knee varus angle | 3.5 | −1.4 | 1.3 | 0.368 |
| Knee flexion angle | 18.7 | 7.4 | 22.0 | 0.121 |
| Knee extension angle | 13.6 | 9.8 | 8.7 | 0.121 |
| Hip extension angle | 9.6 | 7.9 | 8.6 | 0.013* |
| Hip adduction angle | 5.0 | 4.2 | 4.6 | 0.641 |
| Ankle dorsiflexion angle | 21.0 | 22.8 | 20.4 | 0.264 |
| Ankle inversion angle | 2.8 | 4.9 | 7.1 | 0.030* |

The values of moments are presented in Newton meter/kilogram; The values of angles are presented in degrees; differences between stages were compared using the Friedman nonparametric test.

* The significance threshold was was set at 0.05.

This is because the difference in element height between stages 1 and 2 (1.6 mm) was much smaller than the difference in element height between stages 0 and 1 (9.2 mm). Of these parameters, only knee flexion moment and knee varus moment showed an increase in instability from stage 1 to stage 2. This increase was only significant for knee varus moment. More parameters may have shown a significant increase in instability between stages 1 and 2 had the element height at stage 2 been increased. If the element height at stage 2 were increased, then the rise in instability from stage 1 and stage 2 would have been greater. Therefore knee varus angle and hip adduction angle, which do not show a difference between stage 1 and stage 2 in the present study, may have detected such a difference if the height of the stage 2 elements were increased. We believe that this change should be implement in future studies.

These findings suggest that of all the parameters, knee varus moment was the only one that could successfully quantify instability. Of particular interest was the fact that this parameter and all the most sensitive parameters aside from hip adduction moment are measured at the knee. Moreover, the data show that next most sensitive area was the hip, followed by the ankle. This was unexpected since in a previous study we found the distal part of the limb to be most affected by changes in center of pressure (Goryachev et al., 2011). This observation may be explained by the greater translational range of the knee joint as compared to the hip and ankle joint. The difference between the hip and ankle may be due to large movements the hip makes during gait compared to the ankle joint. This gives the hip more opportunity to show variability. Another explanation may be that the distal part of the limb is able to react fastest to instability generated at the foot and therefore STSV was not observed. In fact, a previous study has shown that there is a greater natural variability at the knee and hip in comparison to the ankle (Winter, 1984). Future studies should attempt to generate controlled instability at the waist in order to determine if this observation can be reversed.

These results suggest that a gait analysis of joint kinetic and kinematic parameters would help clinicians evaluate patients who are potentially unstable and at a risk for falling. Clinicians should focus mostly on STSV of the knee joint during motion when evaluating the gait of patients who are potentially unstable. Specifically, the STSV in the kinetic and kinematic parameters of knee flexion moment, knee varus moment, knee flexion angle and hip adduction angle are the most capable at identifying instability. The STSV in knee varus moment is the most sensitive parameter since it can both detect instability and quantify its severity. The results of this study suggest that higher STSV in these parameters predicts greater instability. Therefore a gait analysis of joint kinetics and kinematics can help determine if a person is unstable during gait. Furthermore, potentially unstable individuals, such as the elderly, can be evaluated over time to determine if and at what rate the STSV in these parameters has changed. As a whole, this data can help clinicians decide if a patient is unstable, the severity of the instability, if the instability is worsening over time, if and when to prescribe therapies to combat the instability and if preventative measures should be taken – such as recommending walking with a walking stick – to prevent potential falls. In addition, these findings can also be applied to studies evaluating the success of new therapies for improving stability.

The present study also identified parameters for which the peak magnitudes of the parameter changed significantly with increased instability, aside from the increase in STSV. A significant increase in peak magnitude across stages of instability was found in ankle dorsiflexion moment, hip adduction angle and ankle inversion angle. These changes made be part of the body's adaptation to instability. Adducting the hip and inverting the ankle may keep the body's COM closer to its center, thus minimizing the fluctuation of the COM over the BOS. On the other hand, these findings could also be caused by the device itself. The device changes the biomechanical properties of the lower limb. These changes in moment arms may influence the kinetic and kinematic parameters. A further investigation of these changes by clinicians and researchers is warranted.

There were several limitations to the present study. The first is whether the study design was able to accurately capture all of the most sensitive parameters at detecting instability as well as all the parameters that can quantify instability. Several parameters in the study showed an unexpected, non-significant, decrease from stage 1 to stage 2 of instability. Even more peculiar, however, was that, for many of these parameters, there was a significant difference between stages 0 and 2 and not between stages 0 and 1, even though stage 1 had a greater STSV than stage 2. When the data was looked at more closely it was noted that this result had occurred because of the nonparametric tests used to analyze the data, which examines ranks instead of means. It seems that the distribution of STSV in stage 1 was irregular and often dipped below the values of stage 0 or was skewed otherwise. This is observed clearly in Fig. 3a and c. The STSV in stage 2, however, although sometimes lower on average than in stage 1, were more consistently greater than stage 0. Taken as a whole, we believe this supports our hypothesis that there are increases STSV in kinetic and kinematic parameters when a sufficient level of instability occurs. Nevertheless, it also suggests that the study was not sensitive enough at detecting the difference between stage 0 and stage 1. This may be due to the small cohort of the study. With this considered, future studies may benefit from a larger patient population.

Another option may be to increase the number of trials at each stage of instability. A study by Owings and Grabiner, suggests that the precision of STSV data increases significantly as the number of strides recorded and analyzed is increased (Owings and Grabiner, 2003). While this is difficult to do it a three-dimensional motion analysis gait lab (Hausdorff, 2005), future studies should attempt to carry out many more trials by attaching the motion analysis system to a large treadmill with force plates and ask the patient to walk for one extended trial at each stage of instability. Nevertheless, the findings of the present study are relevant to the many patients undergoing gait analyses for other reasons.

The applicability of these results to patients is also limited. Our schema of instability, although controlled, is different in nature to instability in the elderly or patients with neurological disorders. It is difficult to determine if our schema of instability mimics the cause of instability in these individuals. Unstable gait can present very differently in different individuals who are affected. In some patients, central or peripheral neurological degeneration is responsible for symptoms. In others, instability results from physiological changes – often age-related – affecting the visual, musculoskeletal, vestibulocochlear, somatosensory and cardiovascular systems. It is assumed that the center of mass of individuals who are unstable will increasingly vary in comparison to the base of support (Hollman et al., 2007a,b). The model utilized in the present study mimics gait instability via a perturbative, rocker-type motion during gait. It is important to emphasize that it may not completely overlap with physiological findings in some unstable adults. Nonetheless due to the multi-factorial nature of this pathology, no one model could be suitable for all patients. Our model must be compared in future studies with several types of instability disorders. Regardless, considering almost all the parameters of the present study were able at identifying instability in healthy individuals, it can be assumed that most will detect instability in patients as well.

In addition, participants in the present study walked at a relatively slow walking speed ($1.15 \pm 0.16$ m/s) compared to normal young adults (1.40–1.50 m/s) (Bohannon, 1997). This probably occurred since the subjects walked inherently slower while wearing the apparatus with elements than normally. This slower pace while walking with the device was set to each subject with the metronome before data capture. Therefore the slow walking speed was kept consistent per participant throughout the trials. The slower walking speed in the present study may give a more accurate model of the gait patterns of the elderly and of individuals classified as fallers since they too have been found to walk at slower walking speeds (Callisaya et al., 2010; Toulotte et al., 2006).

It is also important to add that the present study only evaluated the kinetic and kinematic parameters in the sagittal and coronal planes. The kinetic and kinematic parameters in the transverse plane were not considered due to the inaccuracy of the results for these parameters. It was also assumed that these parameters would be least affected by an increase in instability. Nevertheless, future studies should examine these parameters to determine if similar changes occur.

In conclusion, global instability during a dynamic task can be identified using the STSV in kinetic and kinematic parameters of gait when a sufficient level of instability is presented. The most sensitive are knee flexion moment, knee varus moment, knee flexion angle and hip adduction angle. Knee varus moment not only can identify instability but can be used quantify its severity as well. These findings may benefit clinicians evaluating patients who are potentially unstable or for the development of new therapies for treating unstable patients.

## Conflict of interest statement

No author has any conflict of interest to declare.

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## Appendix. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2012.03.007.

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