QUANTIFICATION OF BALANCE IN SINGLE LIMB STANCE USING KINECT

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ABSTRACT

This paper presents a novel single limb body balance analysis system which will aid medical practitioners to analyze crucial factor for fall risk minimization, injury prevention, fitness and rehabilitation programs. We use skeleton data obtained from Microsoft Kinect which captures full human body as well as ensures user's privacy. A new eigen vector based curvature analysis algorithm is developed to compute single limb stance (SLS) duration on the skeleton data. Two parameters vibration-jitter and force per unit mass (FPUM) are derived for each body part to assess postural stability during SLS. Experimental results show the efficacy of our system to apply it in medical domain.

Index Terms— Kinect, Single Limb Stance Exercise, Postural Stability, Balance.

1. INTRODUCTION

Synchronized and coordinated activation of the postural muscles of the trunk and lower limbs is required for maintaining equilibrium and balance in human body. Poor postural balance control causes injury or falls in huge population and is supposed to be a critical factor of common motor skills [1]. Several techniques [2] [3] already exist in the literature for measuring postural control in any stance. Among them Single Limb Stance (SLS) [4] is a good option which not only assesses postural steadiness in a static position by a temporal measurement (SLS-duration) but also analyses the role of body joints in postural stability and correction [5]. For clinicians, it provides a quick, reliable and easy way to screen their patients for fall risks and is easily incorporated into a comprehensive functional evaluation for older adults. The SLS cut-off time for patients suffering with Parkinson's disease is about 10 seconds and it reflects the highest sensitivity and specificity measure for fall-history. SLS training for healthy subjects reduces chances of injury or fall by improving static balance [6]. Moreover, any balance assessing algorithm needs to be tested on healthy subject for validation before applying it on patients. Being a complex mechanism, lack of postural control also creates postural sway during standing e.g. people with low back pain have been observed to have increased postural sway in standing.

Balance in SLS needs to be assessed in terms of both SLS-duration [7] and body-sway which can be measured by center of pressure (COP) movements registered using stabilometry with force platforms [8]. The COP is very much indicative of both the center of gravity's horizontal location and ground reaction forces due to muscular activity but does not inform about how postural perturbation creates instability in different body parts. Eva et al. [9], considered only amplitude of COP movements but omitted frequency associated with each joint vibration whereas clear relationship exists between the oscillation of COP and center of mass (COM). Recently marker-based motion analysis systems [10] had been used for the purpose, but they are obtrusive, expensive and complex. Yang et al. [11] studied reliability of Kinect for assessing the standing balance in terms of COM parameters but they did not discussed about body vibration during SLS in Euclidean coordinate *x,y,z*.

Under this circumstance, in this work we have proposed an automatic unobtrusive system to measure SLS duration and body balance. For this purpose, vibration-jitter analysis is performed which gives a clear view of relative variation of frequency of different joints over time. The whole processing is done on the skeleton data obtained from Kinect. Skeleton data is used instead of video which ensures user privacy concerns. The key contribution in this work are as follows:

1. An eigen vector based curvature point detection method is proposed to calculate SLS duration from noisy skeleton data obtained from Kinect and it is found to be better than standard curvature detection technique [12].
2. The vibration for different body joints are measured in terms of frequency variation i.e. vibration-jitter and force per unit mass (FPUM).

The paper is organized as, in Section 2, SLS balance analysis is presented which includes dataset creation, noise removal, SLS duration measurement and vibration analysis. Section 3 contains the results and discussion followed by concluding remarks in Section 4.

2. METHODOLOGY TO ANALYZE BALANCE IN SINGLE-LIMB-STANCE

Here static single-limb balance assessment [6] is taken into consideration using skeleton data obtained from Kinect. SLS exercise is opted for analyzing static balance of different healthy subjects. The flow diagram of our proposed method is shown in Figure 1.

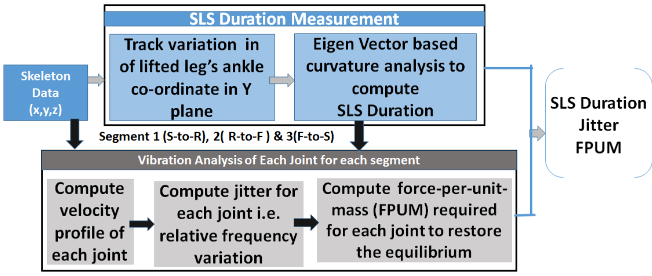


Fig. 1: Block diagram of proposed methodology

2.1. Dataset Creation

As no standard public dataset exists for static single limb balance estimation using skeleton data, we create our own dataset using Kinect [13] [14]. Our experimental setup is shown in Figure 2. Participants perform the tests with bare feet, eyes open, arms on the hips looking straight ahead and standing at 7-8 feet distance away from Kinect. The 3-D spatio-temporal information about 20 joints are obtained from Kinect. For ground truth, time synchronized data capture is carried out using Kinect and Force plate based setup [8]. Subject's video is also recorded to validate our experimental finding manually.

Thirty eight healthy volunteer (age: 21-65 years, weight: 45kg-120kg & height: 4ft6inch-6ft5inch) with no pre-existing symptom of neurological diseases, major orthopedic lesions, vestibular are examined for single limb balance analysis. Three of them did not perform the experiment seriously, so we have discarded their data. Our study is mainly based on the rest 35 subjects. Intentionally we have included few sportsmen (like Subject A & B in Table 1) into our experiment to investigate the effect of physical fitness on single limb balance.

2.2. Noise Removal

Skeleton data obtained from Kinect is very noisy and it is practically visible when the subject stands completely static, but some joints are moving in skeleton. There are many parameters [15] [16] that affect the characteristics and level of noise, which include room lighting, IR interference, quantization noise etc. The noisy skeleton data is filtered using method similar to [17].

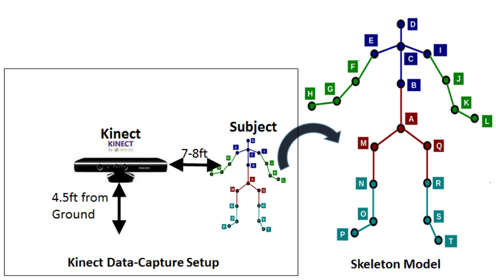


Fig. 2: Data capture setup

2.3. SLS Duration Measurement

During SLS exercise, variation in lifted leg's ankle co-ordinates is very much obvious. We have used this fact in computing SLS duration. The skeleton joints obtained from Kinect are represented by 3D world co-ordinates *(x,y,z)* where 'x' represents left/right variation, 'y' represents up/down variation w.r.t ground and 'z' represents to/from variation of subject w.r.t Kinect. So here, changes in the lifted leg's ankle y-co-ordinate (say, left leg is lifted) $Y_{AnkleLeft}$ can give us meaningful information about the precise timing when a subject lifts leg (here, left-leg) above the ground. Figure 3a clearly legitimizes our claim and shows substantial change in $Y_{AnkleLeft}$ at point R, F and the zone R-to-F is our desired zone of SLS posture. In the other words, R is the frame where foot is flexed off the floor and F is the frame where it again touches the ground. The duration between R & F is considered as SLS duration. Keeping this fact in mind, k-means clustering algorithm is used to capture the variation in $Y_{AnkleLeft}$ with time. It helps us in differentiating one leg stance portion (zone R-to-F). K-means mainly does the segregation (i.e. groups the data into 2 clusters) by optimizing following equation

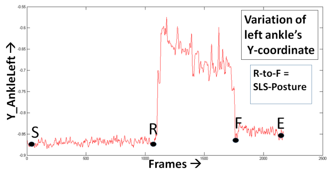
$$O = \sum_{j=1}^{2}\sum_{i=1}^{N} ||Y^{(j)}_{AnkleLeft_i} - c_j||, \; c_j = \frac{1}{N}\sum_{j=1}^{N}(Y^{j}_{AnkleLeft_i})$$

where $||Y^{(j)}_{AnkleLeft_i} - c_j||$ is a Euclidean distance between a data point $Y^{(j)}_{AnkleLeft_i}$ and the cluster center c_j. Frames belong to R-to-F will form one cluster, whereas rest will group into another one, as O is the indicator of the distance of the N data points from their respective cluster centres. Figure 3b shows output of k-means algorithm i.e. frame A and B which are far away from our desired frames R and F. Let us consider data points $\mathbf{X}$ in region S-A. We use the fact that the curvature point will lie in the direction of minimum variance of data. So, we compute covariance matrix $\hat{\mathbf{X}}\hat{\mathbf{X}}^{\mathbf{T}}$ of mean subtracted data $\hat{\mathbf{X}}$ and compute the eigen value decomposition of the matrix. This is the principle behind Principle Component Analysis (PCA) [18] to find the direction of maximum variance. The eigenvector (say, $\vec{E}_{min}$) corresponding to least eigenvalue provides the direction of minimum variance of the data and so reveals the direction towards curvature points. The curvature points R and F are obtained through minimum projection error of the eigen vector corresponding to smallest eigen-value using equation 1.

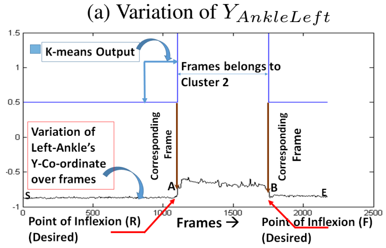
$$\underset{r}{\operatorname{argmin}}[\vec{P}_r - (P_r.\hat{u})\hat{u}] \quad (1)$$

where $\vec{P}_r$ is the original signal value ($Y_{AnkleLeft}(r)$) at frame r (or time instance t); $\hat{u}$ is the unit vector along $\vec{E}_{min}$. Finally

SLS duration is measured by finding difference between time-stamps corresponding R and F frames.



(a) Variation of $Y_{AnkleLeft}$



(b) Curvature points identified by k-means Clustering

Fig. 3: Curvature point detection using variation of $Y_{AnkleLeft}$

2.4. Body Vibration Analysis

Balance is generally defined as person's ability to maintain or restore the equilibrium with minimum movement or sway [19]. Balance is assessed as the amount of postural sway of the human body. Sway is the slight postural movements made by individual joint in order to maintain balanced position.

During SLS exercise while standing on single limb, subject oscillates in order to maintain the balance. Moreover, for a given posture a subject can not move some of the joints like HipCenter, ShoulderCenter etc. easily and flexibly [20]. Hence, the twenty different joints in the skeleton have different degree of freedom (DOF) e.g. it is high for hand but low for HipCenter. This DOF has strong impact on joint movement.

To measure the oscillation quantitatively, velocity profile of each joint is used for its vibration analysis. Vibration is composed of frequency and amplitude. Higher frequency indicates more vibration and less balance. The velocity $\vec{Vel}$ = $[v_x,\ v_y,\ v_z]$ in all the three directions viz. x, y, z is analyzed for estimating the vibration or indirectly balance. The velocity is obtained from the filtered data as following,

$$\vec{Vel} = [v_x^j,\ v_y^j,\ v_z^j] = \frac{[x^j, y^j, z^j]\big|_{t+\delta(t)} - [x^j, y^j, z^j]\big|_t}{\delta(t)} \tag{2}$$

where $[x^j, y^j, z^j]\big|_t$ is the displacement at time t in (x, y, z) direction respectively for j^{th} skeleton joint. It is observed from the AnkleLeft's velocity profile that velocity is maximum near R and minimum (considering sign) near F. Also the the mean velocity of AnkleLeft in first segment S-to-R (see Figure 3a) is almost similar to the the third one, whereas the velocity in the second is much higher than the other two. Start (R) and end (F) frames/time of one leg stance posture have already been identified in the subsection 2.3. Hence, three different segments namely S-to-R (segment-1), R-to-F (segment-2) and F-to-E (segment-3) need to be analyzed separately. This fact is also true for all 20 joints. To get the information about frequency, every joint data in each segment is partitioned into a window of 50 samples and Fourier transform of each segment is evaluated as following:

$$V_k^j(\omega) = \sum_{n=0}^{N-1} v_k^j[n] e^{-i\omega n},\ \ i^2 = -1; \tag{3}$$

where $V_k^j(\omega)$ is the frequency response of i^{th} window for j^{th} joint velocity v_k^j. This is done for all joints and in all three directions *(x,y,z)*. Frequency (f_k^j) corresponding to the maximum amplitude (A_k^j) in each window is selected and the mean frequency of each segment is evaluated as following,

$$f_m^j = \frac{\sum A_k^j f_k^j}{\sum A_k^j} \tag{4}$$

Using above equation 4, mean frequencies $f_m^j\big|_{S-to-F}$, $f_m^j\big|_{R-to-F}$, $f_m^j\big|_{F-to-E}$ in each segment are computed. These calculated mean frequencies will eventually help us to analyze relative frequency variation (vibration) in corresponding segments i.e. before, during and after SLS. In this work, the relative frequency variation is considered as vibration-jitter (in Hz) and for each segment it is mathematically modeled using following equation

$$J_{1,2,3} = (f_m^j - f_{\forall k}^j)\big|_{1,2,3} \tag{5}$$

where $J_{1,2,3}$ is vibration-jitter and f_m^j is mean frequency in each segment whereas $f_{\forall k}^j$ is the frequency for all windows in each segment. $J_{1,2,3}$ also quantifies vibration in terms of frequency for three segments, where more vibration indicates worse balance. For convenience, we will use the term jitter instead of vibration-jitter in rest of the article.

The dominant component of velocity for each window can be written as $v_k^j[n] = A_k^j cos(2\pi f_k^j n)$ where f_k^j is the frequency corresponding to the maximum amplitude A_k^j in k^{th} window for j^{th} joint of each segment.

It is evident from Biomechanics [21] that during SLS, the force imposed on each joint to restore the equilibrium state is due to body weight, abductor muscles force and joint reaction force. This force can be a good measure for joint balance estimation. Keeping these facts in mind, the reaction force per unit mass (FPUM = $\frac{force(F)}{mass(m)}$ in $meter/second^2$) for each joint is measured as the rate of change of velocity for that joint, i.e. acceleration (a). This can be better explained using Newton's law of motion i.e. $F = ma \Rightarrow a = \frac{F}{m}$.

3. RESULTS AND DISCUSSION

This section comprises of the results for SLS duration, jitter and FPUM measurements for estimating total body balance in SLS exercise. We have experimented on 35 subjects and

Subjetcs	Fitness	Segment-1(S-to-R)			Segment-2(R-to-F)			Segment-3(F-to-E)		
		KneeRight	HipCenter	ShoulderRight	KneeRight	HipCenter	ShoulderRight	KneeRight	HipCenter	ShoulderRight
A	10	5.01e-5±2.62e-5	1.53e-6±6.03e-6	9.56e-6±6.03e-6	0.0017±0.0017	2.03e-5±1.89e-5	5.34e-5±4.78e-5	2.26e-4±5.24e-4	7.51e-4±0.0020	0.0012±0.0030
B	9	2.11e-5±9.04e-6	1.59e-5±3.08e-5	1.05e-5±1.21e-5	1.83e-4±1.78e-4	4.21e-5±6.71e-5	1.08e-4±1.65e-4	5.59e-5±1.13e-4	1.32e-4±3.51e-4	2.93e-4±7.52e-4
C	1	2.09e-4±1.25e-4	1.73e-6±1.88e-6	1.36e-5±1.28e-5	7.68e-4±6.00e-4	9.93e-5±1.12e-4	9.38e-4±0.0018	4.19e-4±5.10e-4	1.89e-5±3.13e-5	3.96e-4±5.41e-1
D	2	7.08e-6±4.33e-6	2.80e-6±2.74e-6	6.38e-6±6.21e-6	3.36e-4±4.53e-4	1.18e-4±1.77e-4	0.0016±0.0022	1.53e-5±7.76e-6	2.76e-6±2.65e-6	5.53e-5±5.48e-5

Table 1: FPUM comparison for 4 subjects for three joints. (A,B = sportspersons, but C,D don't practice any kind of exercises)

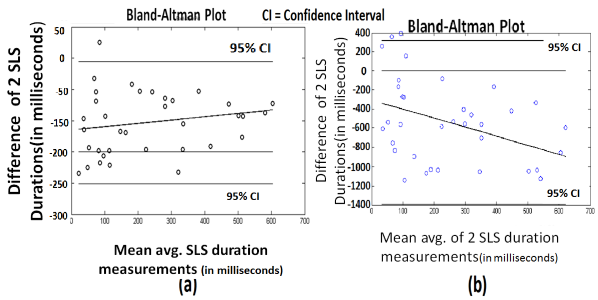


Fig. 4: Comparison of SLS duration for 35 subjects using Bland-Altman plot. (a) Proposed algorithm with GT, (b) [12] vs. GT

have analyzed the results in all three directions, but due to space constraint we are restricting our jitter and FPUM results only in x-direction and reporting FPUM comparison matrix for only 4 subjects in Table 1.

The SLS duration computed from skeleton data using proposed curvature point detection algorithm (mentioned in 2.3) and state-of-the-art technique [12] are compared with the SLS duration obtained from Force Platform based System (Ground Truth/GT) where change in ground reaction force is tracked to get the same. The difference shown using Bland-Altman plot in Figure 4 is mainly plotted against their mean, the mean difference and its 95% confidence levels. Figure 4(a) clearly reveals that the measurements made by proposed method is very much close (max absolute difference 238.3ms) to the ground truth whereas duration computed using [12] is far off from GT (shown in 4(b)). This is due to noisy skeleton data which our eigen vector based curvature analysis algorithm can better handle.

For a particular subject "A", the results for body vibration analysis is shown in Figure 5. As discussed in subsection 2.4, it is quite clear that each and every joint has different order of vibration during SLS. Based on this, for every subject we have analyzed jitter and FPUM (in x-plane) for three joints from upper (ShoulderRight), mid (HipCenter) and lower body (KneeRight). Figure 5(a) clearly depicts that the vibration in ShoulderRight is greater than HipCenter but lower than KneeRight and the observation holds good for all three segments (S-to-R, R-to-F and F-to-E). It is mainly because when body's center of gravity changes during SLS, different body parts having different degree of freedom behave differently to maintain postural stability [21]. For subject "A", the extent of change is high in KneeRight than HipCenter & ShoulderCenter. Figure 5(a) also reflects the same. Similar fact is also verified by observing the recorded SLS video of subject "A".

Moreover, the jitter in segment-1 (S-to-R) and segment-3 (F-to-E) are comparable but much less than segment-2 (R-to-F), as body vibration is more during one leg stance. Although the results are given for three joints in Figure 5, but the same pattern i.e. $J|_{S-to-R} \approx J|_{F-to-E} << J|_{R-to-F}$ is followed by other joints. The above finding is also valid for FPUM based balance analysis as presented in Figure 5(b). The values $mean \pm std$ listed in Table 1 demonstrate how FPUM changes for segment-1, 2, 3 for different subjects and joints. It is also noticed that for every subject either physically fit or unfit, FPUM required to maintain body equilibrium is much in segment-2 than segment-1 and 3. The FPUM value listed in Table 1 for segment-2 is almost 50-100 (e.g. subject 4: 100) times greater than segment-1 & 3 for HipCenter and there is substantial difference for other joints too. However, the results for 4 subjects are presented here but the analysis on several others reveals that the jitter and FPUM values (Table 1) for physically fit subjects is much less than unfit one, which also supports the medical fact in [22].

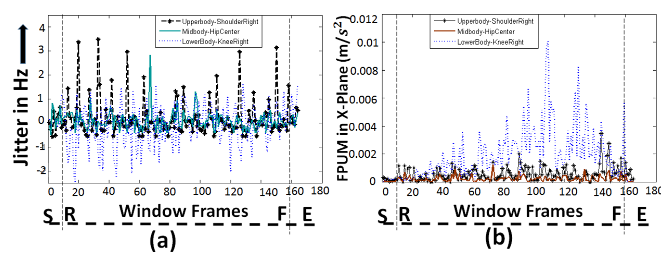


Fig. 5: Body vibration analysis for a particular subject (a) Jitter for different joints for 3 segments, (b) Force-per-unit-mass for different joints for 3 segments

4. CONCLUSION

In this paper quantification of balance using single limb stance (SLS) exercise is proposed and tested on 35 healthy subjects with various fitness level, age, height, weight etc. An effective curvature finding algorithm is proposed which performs better than [12] to find the SLS duration from skeleton data. The quantitative measurement of vibration is formulated in terms of relative frequency variation and force per unit mass (FPUM) for each joint. Results indicate that the vibration jitter and FPUM during one leg stance varies with subjects physical fitness level and is much higher than bipedal one. But how these bio-markers (age, height, BMI etc.) affect single limb balance will be our future scope of research.

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