International Journal of Enhanced Research in Science, Technology & Engineering
ISSN: 2319-7463, Vol. 5 Issue 2, February-2016

Prediction of Human Gait - Vertical Ground Reaction Force

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Abstract: This paper presents a modelling study that helps in mastering gait force pattern generation. Prediction of vertical ground reaction forces one step ahead by using the nonstationary-Markov model of first order is conducted. Examining different gait patterns that differentiate normal from Parkinson introduces two types of gait in which the model is tested. Markov Model is being updated to better fit the ground reaction force signals. It was found that the model better estimate the normal gait and hard to fit different posture segments of Parkinson gait. The ultimate goal is to have a heathy and safe walk to prevent falls among elderly by means of predicting one step ahead. Thus, if the real time measured signal don't follow the path of the estimated data, this could form an indicator of certain perturbation in gait and would serve as an alert.

Keywords: Gait Analysis, Markov model, Nonstationary, Vertical Ground Reaction Forces.

1. INTRODUCTION

Human body requires a continuous contact with the ground during walking and thus ground reaction forces formed as reflections of various forces that the entire body examine during gait. Vertical Ground Reaction Forces (VGRFs) holds the major amplitude and thus has been a topic of interest for many scientists certainly when dealing with its peak value [1]. They are highly correlated to bone growth and strength [2]. Their profile is also proved to examine gait mechanics of powered exoskeleton-assisted walking and reflections to the amount of loading given that subjects are of different level of assist and with different weights and cadences [4]. In addition, VGRFs capture various parameters with no need to measure them. For instance, The peak vertical ground reaction force shows a linear relationship with drop height[1].Furthermore, they are used in diagnosing the effectiveness of surgeries at knee and hip, neuromuscular impairments like in Parkinson's disease, analyses of injury risk, assessment of falling risk, biomechanics and so on [14].

Human walking is produced based on all information acquired from the environment that would affect the walking pattern. With the help of the central nervous system, humans will fit their walk to maximize their stability. Understanding the internal forces inside human body are not limited to rehabilitation, prosthesis design, biomechanics, robotics modelling, sports [1] and others, but exceed to cover sensor validation when it comes to real implementation of gait system. However those forces are hardly measurable [6]. An alternative was to use external forces that are readily observable (experimental data) that serves as an input for musculoskeletal modelling (computer based simulation) to estimate those in vivo forces. This is the potential of inverse dynamic analysis. Having then the VGRF mathematically modelled introduces some inconsistency stemmed from the boundary conditions. For instance, VGRF are modelled exponentially in terms of predefined loading height [1]. However, modelling VGRF mathematically in which parameters are updated based on historical values is introduced as an alternative method. This would help in avoiding in estimating new parameters. It has been tested and validated. The first half samples of observations are used for parameter estimation and the other half are used for validation. An estimate of one step signal a head is conducted and the percentage of error is computed. A comparison between normal and Parkinson is then examined. Results show that the model chosen is best fit for normal gait forecasting.

This model don't cover in one equation both distinctive phases of gait: stance and swing. In fact, a switch control system is proposed that switches between the two phases of gait that corresponds to the real practiced segment during walking. This paper focuses on the stance phase.

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2. MATERIALS , METHODS& OBSERVATIONS

A. Database

VGRFs in Newton as a function of time are extracted from 8 sensors (Ultraflex Computer DuynoGraphy, Infotronic Inc.) underneath each of the right and left foot. Each subject walked at his/her usual back and forth for two minutes at their self-selected pace level ground without any secondary task in a well-lit, obstacle free, 25-m long, 2-m wide corridor. They are captured from 93 patients with idiopathic PD (disease stage was2–3 on the Hoehn and Yahr scale, mean age: 66.3 years; 63% men), and 73 healthy controls (mean age: 66.3 years; 55% men). Subjects provided written informed consent prior to performing the experiment. The database of VGRF Data were drawn from physionet database [7]. The sensors' location inside the insole as lying approximately at the following (X, Y) coordinates measured as a person is comfortably standing with both legs parallel to each other as shown in Fig.1. The origin (0, 0) is just between the legs and the person is facing towards the positive side of the Y axis.

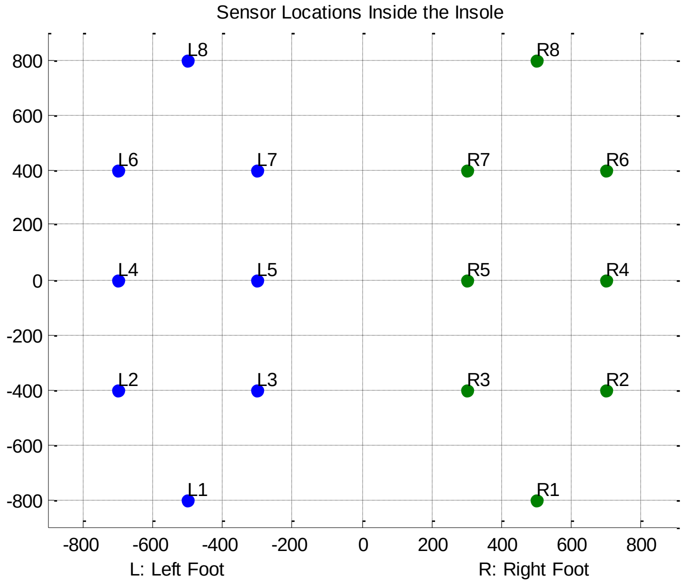

Figure 1. Sensor's position as distributed underneath both feet.

The sampling rate is 100 Hz. Fig.2 displays a sample of the data captured by the array of the eight sensors in addition to their summation. Normal and Parkinson gaits are presented.

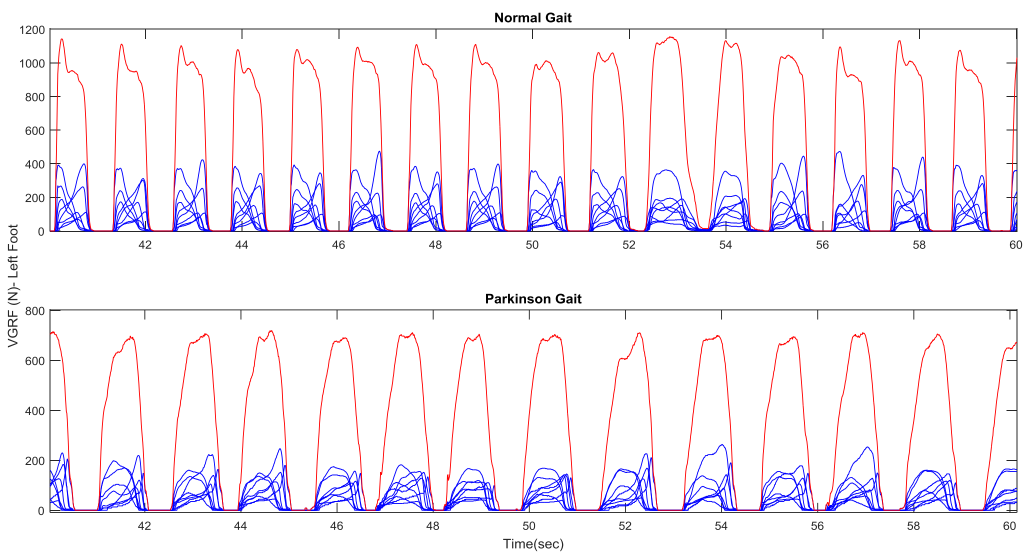

Figure 2. Sample of VGRF data captured by the eight sensors underneath the left foot. The red curve represents their summation. The first row corresponds to normal gait and second raw corresponds to Parkinson gait subject

B. Periodizing VGRF Signals

On-off controller simply drives the acquisition of VGRF during walking from fully closed to fully open depending on the location of the foot. In other words, when the foot hits the ground entering the stance phasethen VGRFs signal do exist. The acquired data will be saved into an array with preliminary fixed size ending up with a matrix of stance phases. Its size is controlled by the preliminary defined stance interval depending on the sampling rate. This would eliminate any difference in the stride interval between subjects stemmed from their difference in height, weight, gender and so on. The signal then have fixed time periods (i.e. the moment from heel strike into the other is equal from another heel strike into another). That's why the VGRF signals are divided as shown in Fig.3.

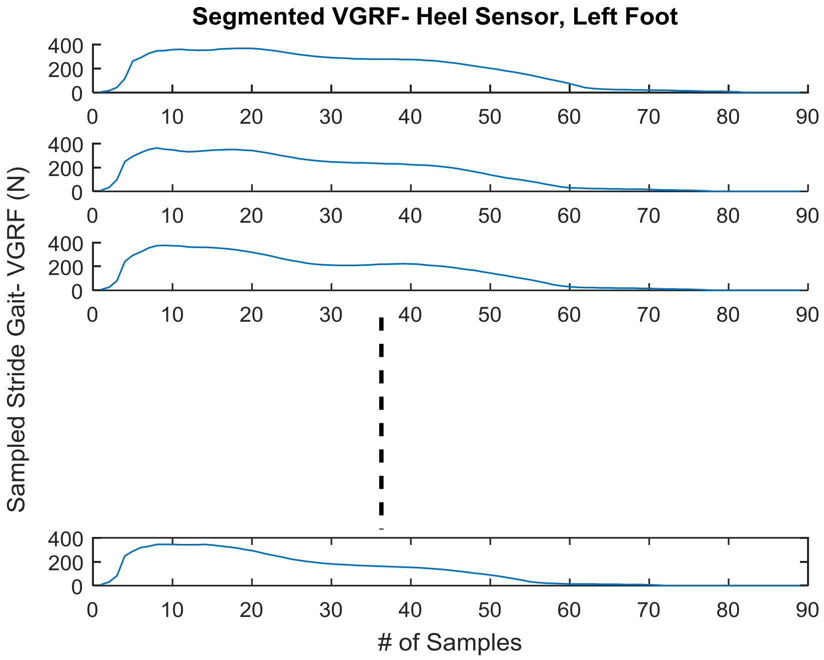

Figure 3. Step Isolation of VGRF stance phases at the heel. They are saved over 89 samples equivalent to 0.89 seconds.

The matrix of stance phases are reshaped into 1D array vector for analysis. Such fixing of sampling points interval for stance phase of the gait would definitely change frequency content analysis. Definitely, they will exist at different harmonics of the saved interval as shown in Fig.4:

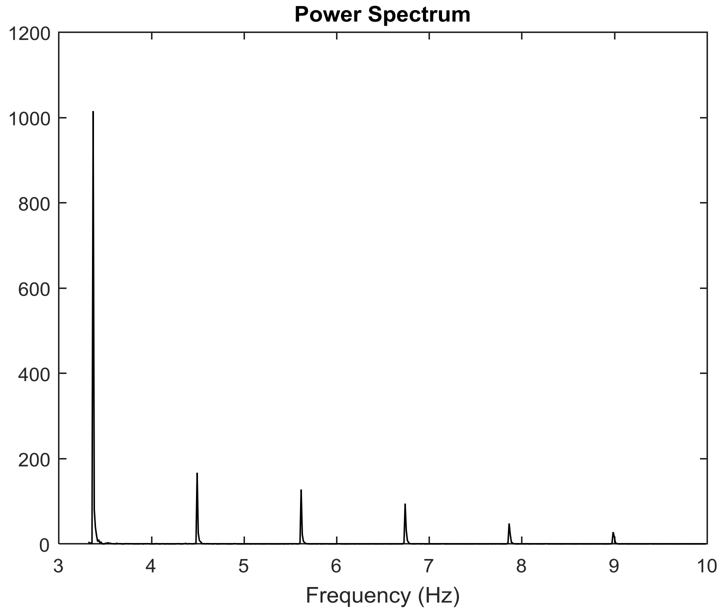

Figure 4. Power Spectrum of the new generated signal.

The Autocorrelation Function (ACF) of the altered signal indicates a very slow decay over lags and thus the signal becomes more non-stationary. This is inherited from the periodicity being added to the signal. Fig.5 is an illustration.

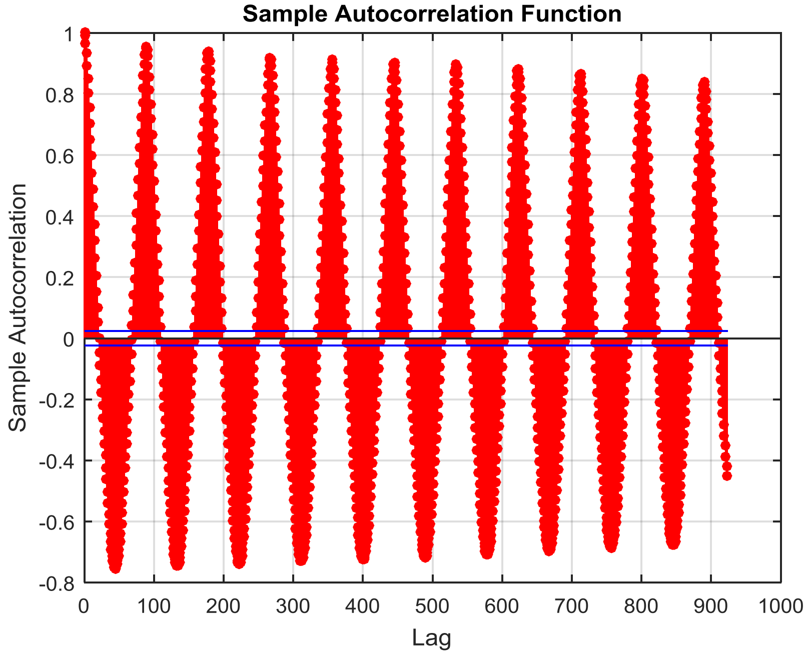

Figure 5. Sample Autocorrelation Function

Differencing the new signal by two consecutive times as an intention to remove periodicities signifies the interval chosen by implementing Partial Autocorrelation Function (PACF) as shown in Fig.6. It agree with the predefined interval of 89 lags:

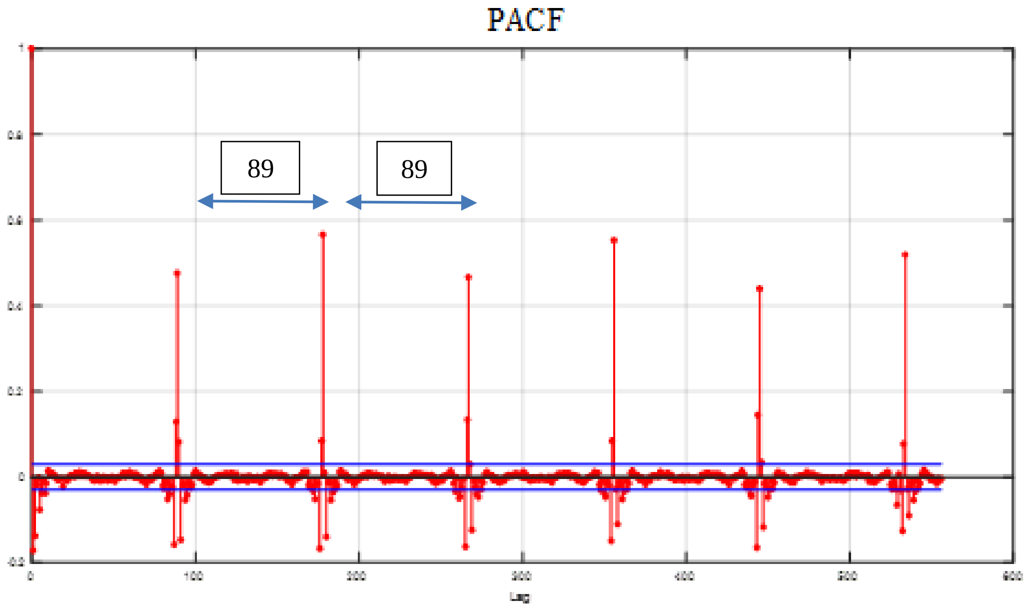

Figure 6. Partial Autocorrelation Function

C. Time series Comparison

Fig.7 indicates that the extracted stance phases from a normal subject VGRFs gait delivered by the heel sensor at the left foot preserves some fixed magnitude. This similarlyremark them to be below the sample 20 and therefore below the 0.2 sec. while the Parkinson subject admits some variations certainly in the amplitude of various steps, this can be observed from the peaks in Fig.8. Furthermore the peaks are within the samples of 20 and above. Such a difference give an intuition of two important different properties: the change in the VGRF amplitude from one step into another admitted by Parkinson gait, and a change in time interval to reach maximum amplitude as an evidence to a change in the slope and those the stance phasesare longer in Parkinson gait.

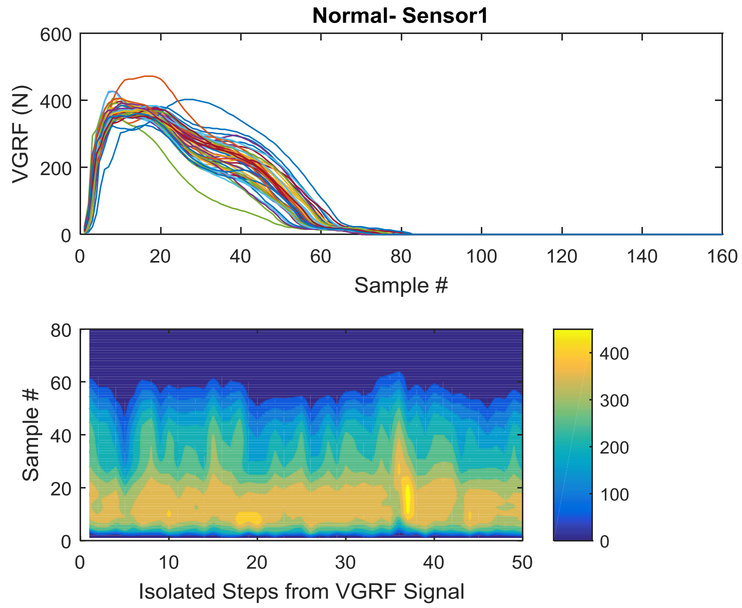

Figure 7. Stance phases being isolated from VGRF extracted from the heel sensor underneath the left foot of a normal gait subject and their contour plot.

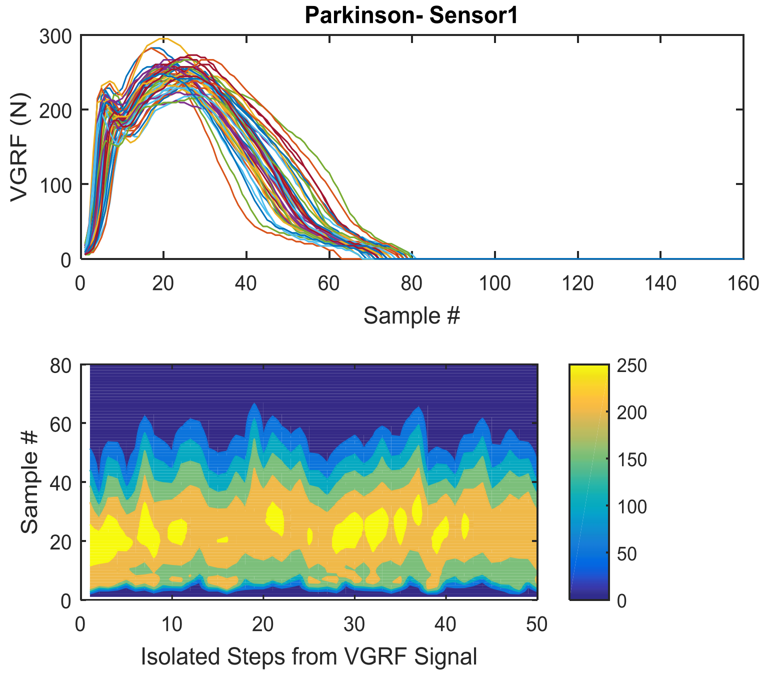

Figure 8. Stance phases being isolated from VGRF extracted from the heel sensor underneath the left foot of a Parkinson gait subject and their contour plot

D. Periodogram Comparison

Now fixing the size to 160 samples by zero- padding the step signalsas to have a size of 160. 160 is chosen as a guarantee that neither of subjects will have stance interval beyond the 1.6 sec during walking. This definitely will create an artificial fundamental frequency of f_f= 0.625 Hzin its periodogram indicating that the cyclic existence of the impulse starting of the gait stance phase. Furthermore a harmonics periodic of the fundamental will also be created (2 f_f, 3 f_f, 4 f_f...). To enhance the outcomes of the comparison, each step extracted from VGRF is normalized by Euclidean norm of the same step signal. This is given in equation (1).

$$||x||_2 = \sqrt{\sum_{n=0}^{N-1} ||x[n]||^2} \tag{1}$$

Excluding the first harmonic, the amplitude of the power spectrum in Normal subjects is higher than a Parkinson subject as indicated in Fig.9, this is a good indicator that slope of VGRF measured at a certain sensor during the moments of contacting the ground is steeper than the one known by a Parkinson.

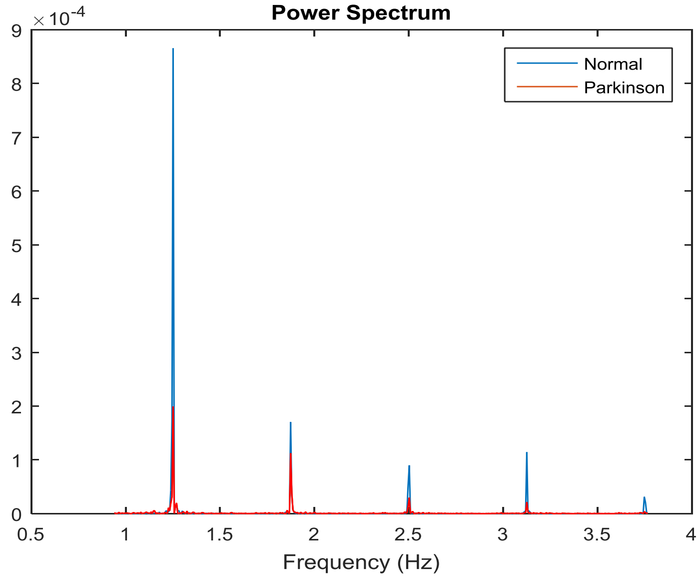

Figure 9. Power Spectrum for both altered signals of Normal and Parkison gait subject

E. Hypothesis

Hypothesis: One stance VGRFs gait is directly affected by the very neighbor steps and up to a certain interval (i.e. number of steps). The above hypothesis doesn't indicate whether one step affect the whole style of walking but definitely get capture some characteristics of the previous step and affect the step gait a head. The accumulation of errors between two steps that are far away from each other contribute also ingait variability. This means that certain parameters are needed to be adjusted continuously while generating the upcoming signals. To realize this hypothesis in practice, a model will be chosen then the parameters are fit into equation to achieve the minimum error between actual and simulated data. Then, modelling vertical ground reaction force signals based on historical data can reproduce and forecast the gait signals for a short period of time.

F. First order Markov model

Suppose the future step values to be predicted are generated by the following 1st order Markov Model that is stationary with respect to mean, variance in addition to auto-correlation of first lag, this is indicated by equation (2).

$$X_{j+1} = \mu_x + \rho_1 (X_j - \mu_x) + t_{j+1} \sigma_x \sqrt{1 - \rho_1^2} \quad (2)$$

Where t_{j+1} stands for the standard normal variable (~ N (0, 1)) which is a series generated randomly with zero mean and unity variance. μ_x, σ_x and ρ_1 are the mean, standard deviation and first lag autocorrelation respectively.The same model will be generalized for nonstationary process as indicated in equation (3).

$$X_{i,j+1} = \mu_{j+1} + \rho_j \frac{\sigma_{j+1}}{\sigma_j} (X_{ij} - \mu_j) + t_{i,j+1} \sigma_{j+1} \sqrt{1 - \rho_j^2} \quad (3)$$

Where i is the number of stance phase of gait step, ρ_j is serial correlation between the jth moment of a stance phase and j+1th moment of the same stance phase. Once again the standard variable series is given by $t_{i,\, j+1}$~N(0, 1).

3. RESULTS

G. Nonstationarity of VGRF

Different Models could be adapted for VGRF for data prediction and forecasting. Once time series shows a constant mean and variance besides to when autocorrelation only depends on time lag, the signal is supposed to be stationary. However, VGRF is said to be non-stationary signal [8] and cyclostationary (non-stationary that does not include a trend-like behavior) as in reference [9]. In this section, the approach monitored from wide point of view by assuming VGRF data to be non-stationary. This is verified by taking the mean and the standard deviation of segments that corresponds to a given step. Fig.10 presents those statistics to vary from one step into another. The mean ranges from 99 to 200 N and standard deviation ranges from 117 to 174.

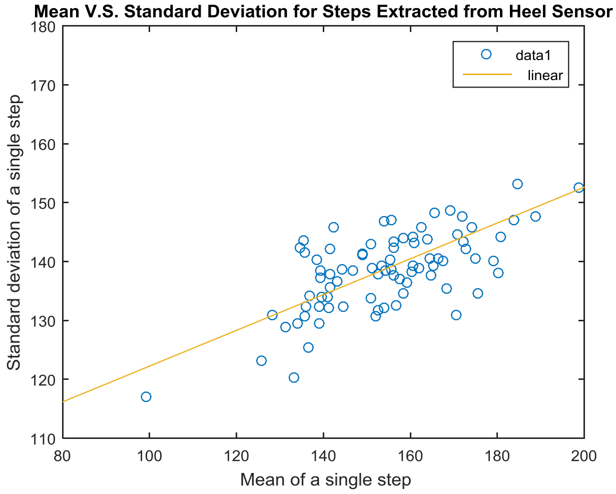

Figure 10. Standard deviation versus mean for a normal subject upon Heel sensor located in the left foot.

This forms a good indicator that the signal is non-stationarity certainly in the variance and in the mean. However, this is not to contradict for the moment the cylostationarity hypothesis of the signal that is a form of non-stationarity [5]. In our case the mean and standard deviation are highly correlated. In consequence, a lognormal model is greatly useful in modelling.

H. First order Markov model with non-stationarity

The initial guess of parameter values in gait VGRFare taken to be zero and this makes sense in the real gait that every stance phase is followed by a swing phase where there is no contact with the ground and therefore the VGRF measured must be zero.

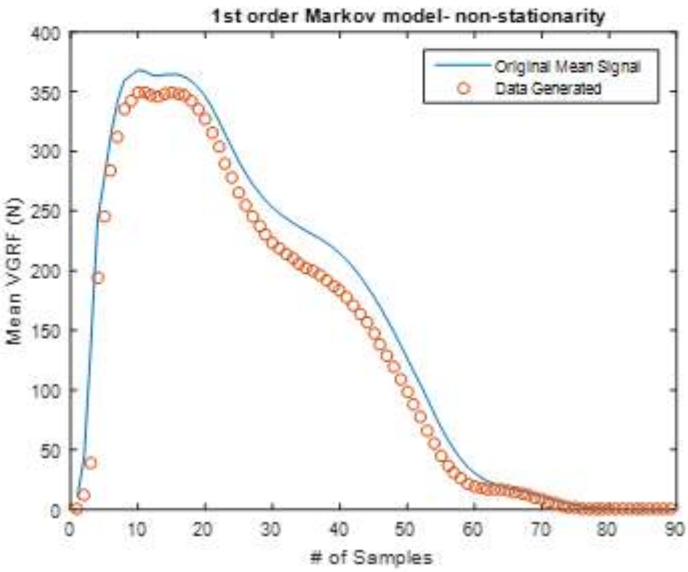

Figure 11. Data generated by markov-Model V.S. Raw Experimental Data

The coefficient of multiple determination (R-square) is a good choice for testing the model as the outcome of the square of the correlation between actual values and predicted values, would indicate a proportion of variance that is accounted by the model. This can be articulatedin equation (4).

$$R^2 = \frac{\sum_{i=1}^{n} w_i(\hat{y}_i - \bar{y})^2}{\sum_{i=1}^{n} w_i(y_i - \bar{y})^2} \qquad (4)$$

Where $\hat{y}_i$is the fitted values and $\bar{y}$ represents the mean. Finding R^2 for the model in Fig.11. for a normal gait subject yields a value of 0.9143 and this means that the fit explains 91.43% of the total variation in the data about the average.

I. Modified First order Markov model with non-stationarity

The first step in modifying the model, is by changing the correlation between previous moment of the step and the second moment into autocorrelation of the previous moment only. Then as the above model is restricted to certain random values and falls under a standard normal distribution, it can best be adjusted by replacing it with the error term ending up with equation (5).

$$X_{i,j+1} = \mu_{j+1} + \rho_j \frac{\sigma_{j+1}}{\sigma_j}(X_{ij} - \mu_j) + (1 - R^2)\sigma_{j+1}\sqrt{1 - \rho_j^2} \quad (5)$$

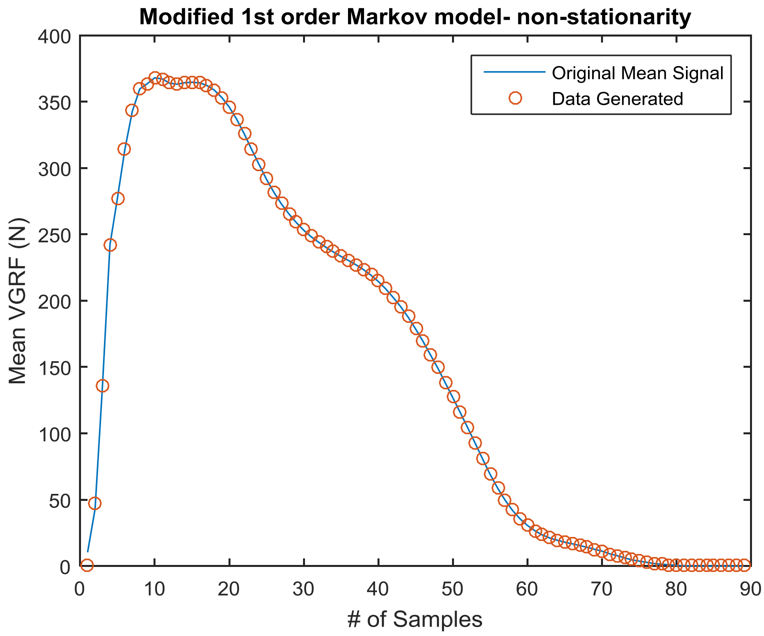

Figure 12. Data generated by the modified Markov-Model V.S. Raw Experimental Data

An acceptable data prediction as shown in Fig.12 is then reached using the new model. The percentage of fitting increased to reach 99.76% on average.This result is generalized to cover all other sensors with variable percent of error but still are highly acceptable. A summary of the algorithm is expressed in Fig.13.

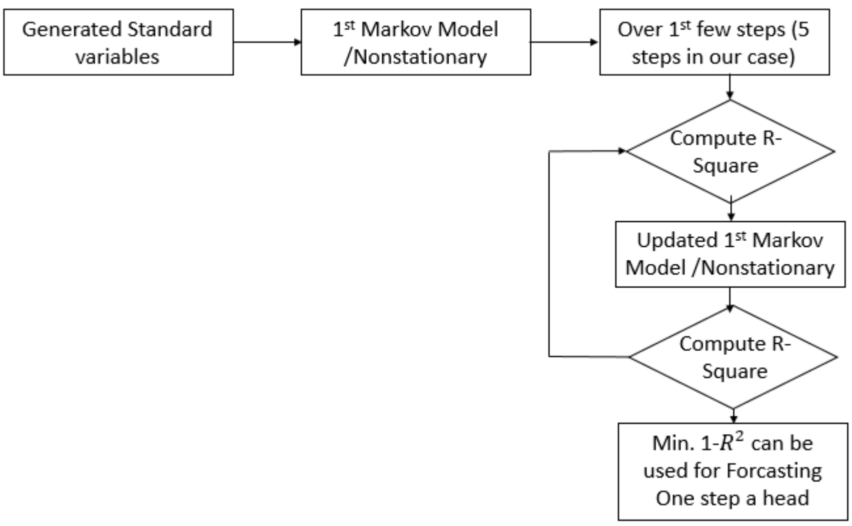

Figure 13. Algorithim for Modified Markov Model

J. Prediction of one step ahead in "Normal" & Parkinson"

After assessing time domain statistical properties of the Vertical Ground Reaction Force (VGRF) during moderate-pace walking, the aim is then eventually to create a reliable mathematical model of VGRF for normal and abnormal cases and that what have achieved so far. Predicting a one-step signal ahead in normal and Parkinson using the above model yields an important difference between them. The proposed model is able to predict in normal subjects better than the Parkinson subjects as shown in the Fig.14.

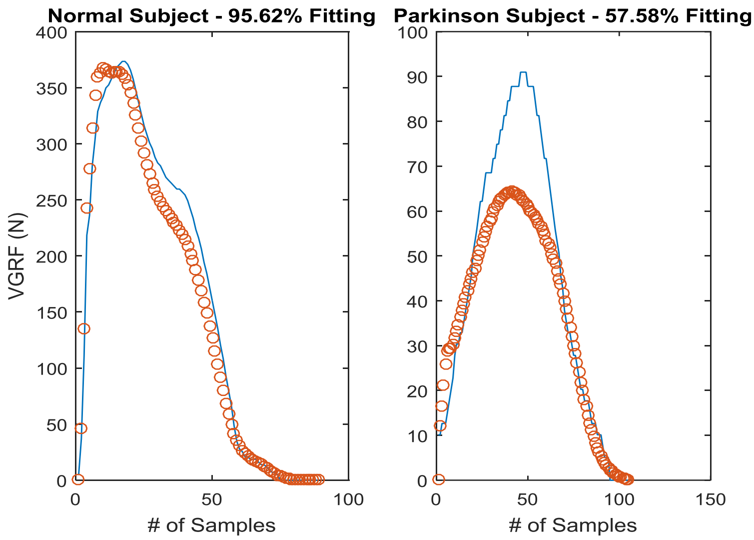

Figure 14. Forcasting Normal and Parkinson Gaits.

Fig.14 shows a 95.62 percentage of fit in the predicted normal gait signals while the model experienced a hard moments in having an appropriate prediction in Parkinson gait where the fit is around 58%. This can be justified by the previous observation in Fig.8 where the gait signals in Parkinson experience great variability.

4. DISCUSSION & CONCLUSION

Autoregressive Moving Average (ARMA (5, 5)) linear model is conducted on VGRF [10]. However this technique is fit for stationary time series signals and therefore periodicities are not desirable. It is worthy that VGRF are assumed to be cyclostationary [9] and nonstationary as on this paper. Furthermore, the several combination of orders must be conducted then the best model will be chosen based on verification tests like Maximum likelihood rule. In another research the linear and polynomial regression models are being used and tested by Normalized Root Mean Square Error. However, the linear model didn't capture the peak forces and the polynomial didn't capture the dynamic patterns in force profiles. In order to increase estimation accuracy, the writer suggests to have different fitting methods for better explaining of the unique patterns in force profiles, and thus [11]. Another work on modelling VGRFs signals is introduced by the fourth order polynomial with additional two parameters. However, this technique requires the location of the local maxima in the actual VGRF data [12]. Therefore, the model will capture a huge error when both feet are in contact with the ground (i.e. no aerial phase). That is why, building models based on historical data is helpful specifically when it comes also to the type of foot: normal, low, and high-arch foot [4]. Research shows that this would affect the pattern of VGRF [13] making it difficult to have a common model generalized over different human gaits. This paper presented a novel modelling technique of VGRF signals that overcome the stated problems and preserves the fact being nonstationary. The technique is summarized on a modified first order Markov model based on nonstationary characteristics. It has been evaluated on normal gait signal and proved its potential in having a good prediction. However, it is hardly estimated the Parkinson gaits which suggests a higher order must be implemented. As a future work, the model will be verified over all sensors by finding a unique multi-sensor array model. On the other hand, the model generated should take into consideration the time quantization of the gait and thus the stride interval and also adopt the lognormal model into the proposed model based on the mentioned observation.

REFERENCES

[1]. W. Niu, T. Feng, C. Jiang and M. Zhang, "Peak Vertical Ground Reaction Force during Two-Leg Landing: A Systematic Review and Mathematical Modeling," BioMed Research International, Hindawi, vol. 2010, p. 10 pages, 2014.

[2]. A. Pouliot-Laforte, L.-N. Veilleux, F. Rauch and M. Lemay, "Validity of an accelerometer as a vertical ground reaction force measuring device in healthy children and adolescents and in children and adolescents with osteogenesis imperfecta type I," J Musculoskelet Neuronal Interact , vol. 14, no. 2, pp. 155-161, 2014.

[3]. D. B. Fineberg, P. Asselin, N. Y. Harel, I. Agranova-Breyter, S. D. Kornfeld, W. A. Bauman and A. M. Spungen, "Vertical ground reaction force-based analysis of powered exoskeleton-assisted walking in persons with motor-complete paraplegia," J Spinal Cord Med. , vol. 36, no. 4, p. 313–321, 2013 .

[4]. S. Boozari, A. A. Jamshidi, M. Ali Sanjari and H. Jafari, "Effect of Functional Fatigue on Vertical Ground-Reaction Force in Individuals With Flat Feet," *Journal of Sport Rehabilitation, Human Kinetics, Inc.*, vol. 22, pp. 177-183, 2013.

[5]. R. Alkhatib, C. Corbier, M. El Badawi, B. Moslem, M. Diab,"Sensor's Ground Reaction Force Behaviour for Both Normal and Parkinson Subjects - a Qualitative Study", 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Societ. Mico-Milan Conference Center – Milan, Italy, August 25-29 2015.

[6]. D. S. Bloswick and D. B. Chaffin, "AN ERGONOMIC ANALYSIS OF THE LADDER CLIMBING ACTIVITY," International Journal of Industrial Ergonomics, Elsevier, vol. 6, pp. 17-27 , 1990.

[7]. M. Jeffry. F. Research, "Gait in Parkinson's Disease," National Institutes of Health, National Parkinson Foundation, and the Parkinson's Disease Foundation , Physionet, 2005.

[8]. Alkhatib, R., Diab, M., Moslem, B., Corbier, C., & ElBadaoui, M. "Classification of Ground Reaction Forces Based on Non Stationary Analysis", Proceedings of the International Conference on Computer Vision and Image Analysis, IEEE ICCVIA 2015, pp. 1-4 , Soussa, Tunisia, Jan. 2015.

[9]. K.Sabri, M.ElBadaoui, F.Guillet, A.Belli, G.Millet, J.Morin, Cyclostationary Modeling of Ground Reaction Force Signals, Signal Processing, Vol. 90, No. pp.1146-1152, 2010.

[10]. J. Garza-Ulloa, "Sensor Validation using Linear Regression for Error Detection between prediction data behavior and acquired data applied on sensing vertical Ground Reaction Forces on Human Gait Analysis.," International Test and Evaluation Association (ITEA) White Sands Missile Range, pp. 24-27 , 2012.

[11]. S. Lee and T. J. Armstrong, "CPWR Final Report," in Field Tool for On-Site Biomechanical Analysis During Ladder Climbing, University of Michigan , The Center for Construction Research and Training, 2014, pp. 13-14.I. S. Jacobs and C. P. Bean, "Fine particles, thin films and exchange anisotropy," in Magnetism, vol. III, G. T. Rado and H. Suhl, Eds. New York: Academic, 1963, pp. 271–350.

[12]. T.Anthony, E.W.Roshana, "Biomecahnics of Running and Walking, Issue 43 of Dolciani Mathematical Expositions," in *Mathematics and Sports*, contibutor:Mathematical Association of America, MAA, 2010, 2010, p. 324.

[13]. Y. Yorozu, M. Hirano, K. Oka, and Y. Tagawa, "Electron spectroscopy studies on magneto-optical media and plastic substrate interface," IEEE Transl. J. Magn. Japan, vol. 2, pp. 740–741, August 1987 [Digests 9th Annual Conf. Magnetics Japan, p. 301, 1982].