Frequency Content Analysis of Gait -Vertical Ground Reaction Force

Hiba Karout, Mohamad Diab, Bassam Moslem
Electrical and Computer Engineering Department
Rafik Hariri University
Damour, Chouf 2010, Lebanon
AlkaroutHM@students.rhu.edu.lb, diabmo@rhu.edu.lb, moslembm@rhu.edu.lb

Rami Alkhatib, Christophe Corbier , Mohamed El Badaoui
Laboratoire LASPI, Université Jean-Monnet, 20 avenue de Paris, 42300 Roanne, France
rami.alkhatib@universite-lyon.fr, christophe.corbier@univ-st-etienne.fr, mohamed.elbadaoui@univ-st-etienne.fr

***Abstract*—this paper addresses the frequency content of Vertical Ground Reaction force (VGRF) signals. Signal processing of such data is crucial in order to characterize the gait itself. The outcome processing can be used to show the presence of disease, to track the rehabilitation or even to detect the possibility of falling. In order to come up with the correct processing, it is important to set up the rules that cover gait VGRF. In this paper, an attention is given to the frequency content of the VGRF underneath different locations of the foot. The extracted information is then used to compare gait between normal and Parkinson's subjects. Results have shown the existence of three dominant frequencies. Furthermore, it was revealed that the power of frequency is high for the second peak in normal gait compared to Parkinson.**

Keywords—Frequency analysis, Synchrosqueezed time-frequency;

I. Introduction

Up to 1.2 million of Parkinson's patients were recorded in Europe according to National Parkinson's Foundation [1]. Vertical Ground reaction force (VGRF) signals have always been a point of interest to many researches [2]. VGRF is the force exerted by the ground on a body in contact with it in the y-direction. In this study, we analyze the frequency content of two types of subjects: subjects with Parkinson's disease and normal subjects.

Several techniques could be used to transfer a signal from time to frequency domain such as FFT [3]. Antonsson's database composed of two categories, runners and walkers. He studied 30 foot contacts from 12 subjects. By applying FFT spectral power analysis of these force records, the results showed that for walking, 99% of the integrated power content of VGRF signals below 9 Hz. For running, 99% of the integrated power content of the VGRF signals was at frequencies less than 10 Hz [3,4]. Furthermore, the amplitudes above 10 Hz are recorded to be less than 5 % of the fundemetal. 2 % above 20 Hz and all amplitudes more than 1 % are delimited below 50 Hz. 99.5 % frequency in VGRF is recorded to be 6.39±2.31 Hz in Parkinson subjects. The median frequency is found 0.45±0.09 Hz and the bandwidth is said to be 1.23±0.29 Hz. A difference is found in frequency content when compared to healthy subjects [5]. After averaging the VGRF signals of PD subjects, the power of high frequency is lower and the first and second peak's amplitude are lower than normal subjects with a delayed occurrence of the first peak. The average power is between 0.5 Hz and 1.5 Hz is logged in PD [6].

These results helped in the process of selecting the appropriate filter that is the fourth order Butterworth filter with a 25Hz cutoff frequency. This filter appeared to eliminate 99% of the noise while retaining all of the important components of the signal [4]. Another research done in the University of Nebrashka Omaha has discussed the effect of multiple sclerosis on the frequency content in VGRF during walking by applying Fourier transform on the signals. Wurdeman et al. results showed that patients with multiple sclerosis had significantly lower than 99.5% frequency (P= 0.006) and median frequency (P<0.001) in the vertical ground reaction force where P is the value of significance . The lower frequency content suggests lesser vertical oscillation of the center of gravity [7]. Analysis of the frequency content may potentially serve to provide earlier diagnostic assessment of this debilitating disease [7].

Likewise, frequency content of various body movement is recorded to be less than 20 Hz where 99% of the FFT spectrum is contained below 15Hz in walking signals [4]. However as a majority of the signal existed below 10 Hz as shown by Simon et al, at the heel contact a peak appeared with frequency content of 10-75 Hz. Giakis and Baltzopoulos indicated that 95 % of signal's amplitude is within the first 15 harmonics up to 20.27 Hz [8]. Vertical GRF impact peak is assumed within 8-50 Hz given that it will visually absent in time domain for forefoot running and a suggestion of an overlap between impact and active peak frequencies do exist [9]. Moreover, the standards for testing prosthetic components required for ankle-foot in order to study the fatigue process are by applying alternating forces within 0.5-3Hz as given by the international organization for standardization. Those values based on conference paper as designated by [10].

The aim of this paper is to further investigate the main fundamentals frequency content of gait VGRF. In addition, a comparison between normal and Parkinson subjects in terms of their frequency is developed.

II. METHODOLOGY

Fourier transform (FT) is used to examine the frequency content of the signal while short time Fourier transform (STFT), wavelet transform, Hilbert transform and many others provide the time-frequency representation of the signal [16]. Depending on the type of analysis and assumptions made, a transform is used. That is why in order to comprehend the frequency content of the VGRF signals over time, both STFT and WT can serve a good foundation, prearranging the linearity and nonlinearity of used signals are not topic of this paper.

The STFT or Gabor transform, $G_s(w,t)$, and its algorithm is given by (1) of signal s(t) [17]:

$$G_s(w,t) = \int s(u)g(u-t)e^{-iwu}du \qquad (1)$$

Where s(u) is the signal of interest, g(u) is the windowed function and "w" and "t" stands for frequency and time respectively[17]. In this paper the Gaussian window is being a topic of interest given by (2). f_0 is the resolution frequency :

$$g(u) = \frac{1}{\sqrt{2\pi f_0}} e^{-u^2 / 2 f_0^2} \qquad (2)$$

The wavelet transform $W_s(w,t)$ is given on (3):

$$W_s(w,t) = \int s(u)\psi^*(\frac{u-t}{a})\frac{du}{a} \qquad (3)$$

"a" is the scale and $\psi^*(u)$ is the chosen wavelet function and is given as lognormal analogous to Gaussian in STFT [17]:

$$\psi^*(f) = e^{-(2\pi f_0 \log f)^2/2} \qquad (4)$$

In order to have more concentrated time –frequency representation, the algorithm of synchrosqueezing of STFT or WT in [17] will be convenient to be used by joining all coefficients having same instantaneous frequency into one coefficient.

III. DATABASE AND PREPROCESSING

A. Database

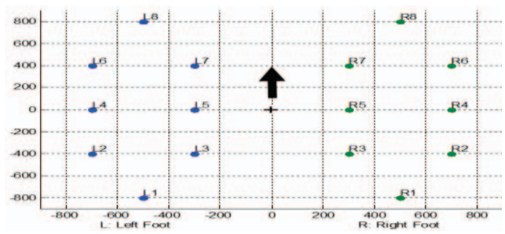

Fig.1: Sensor's position as distributed underneath both feet.

VGRF Data was drawn from physionet database [18]. 18 normal control and 29 Parkinson subjects are used. Eight sensors were placed underneath each of the subject's feet to collect VGRF data for 2 minutes during subjects walk. The sensors are distributed as shown in Fig.1.

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.

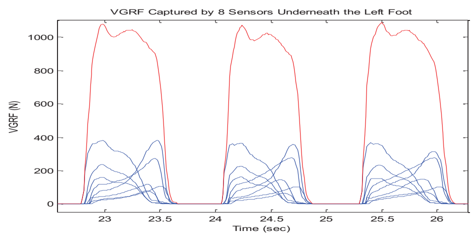

Fig.2: Sample of VGRF data captured by the eight sensors underneath the left foot. The red curve represents their summation

B. Preprocessing

As in the analysis of any acquired signal, its important to pay attention towards the experimenting phases and the enviromental conditions while obtaining VGRF data. Fig.2 indicates ellipses at certain instant of times. When recording the signals, subjects were asked to walk at their own pace for two minutes. Those ellipses are obtained when the subject reaches an end point and then asked / required to turn around. Such turning points as shown in the figure will alter the frequecny content of the signals and therefore the analysis.The frequency at those instants scatters from 0.6 to 0.9 Hz. That's why it is become important to reconstruct those part or the signal have to be segmented.

In order to relate this phenomena to the real experiment, participants walked without any secondary task in a well-lit, obstacle free, 25-m long, 2-m wide corridor [12]. This suggest the change in the frequecny shape over time.

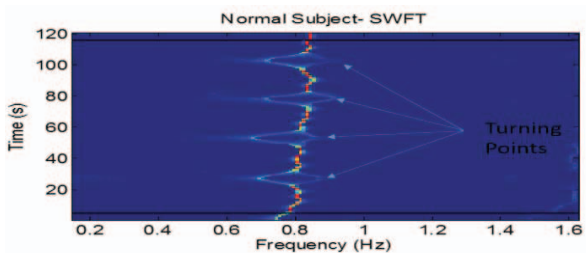

Fig. 3. synchrosqueezed short time fourier transform

IV. RESULTS AND DISCUSSION

A. System's Frequency Nonlinearity

A linear system in frequency is achieved when the net amplitude frequency (Afreq) of the summed signal (total VGRF) is equal to the sum of the amplitude frequencies produced by each sensor's signal. More clearly this is emphasized by (2) where "i" is the sensor number:

$$Afreq(\sum VGRF_i) = \sum Afreq(VGRF_i) \qquad (2)$$

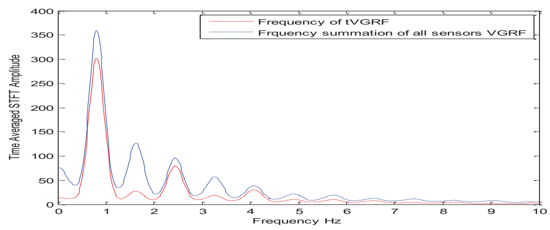

Fig.4: The power of frequency content present in tVGRF and the summed amplitude is different from the summation of frequencies for sensor's VGRF.

However, this is not the case on gait VGRF. Fig.4 emphasizes clearly the difference between the frequency content of the total VGRF (tVGRF) and the summation of frequencies of the array of 8 sensor's VGRF. The most fact is that the second peak becomes of lower amplitude compared to the third peak in tVGRF. This suggests the Heterodyne phenomena around ~ 1.63 Hz. In addition, the even harmonics are most likely to be affected.

B. Frequency content

Up to our knowledge, the marginal level of frequency in the gait VGRF is noted to be less than 20 Hz [12]. Consequently, our focus will not exceed the 20 Hz for the reason that STFT confirms no valuable power frequency content revealed for frequency above 15 Hz. Fig.5 Displays up to 5Hz for graph clarity in addition that concentrated power frequency content is located at this region.

First, all sensors contain the same frequency content with a relatively small different amplitudes as shown in Fig.6 within the same person, this is found in all subjects and can be explained as a tend to maintain coherence in gait at least during experimenting for small duration. This also quantifies that participants are walking on free obstacle level ground

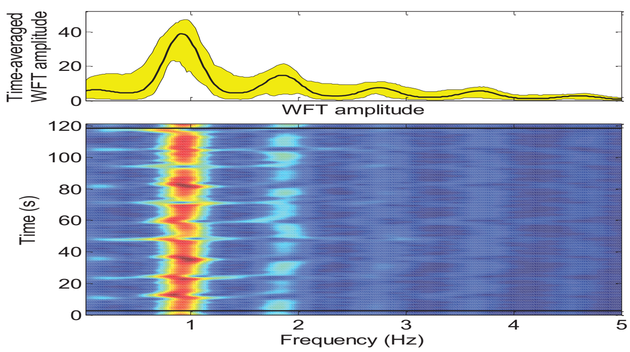

Fig.5: STFT plot and upper plot is time averaged STFT

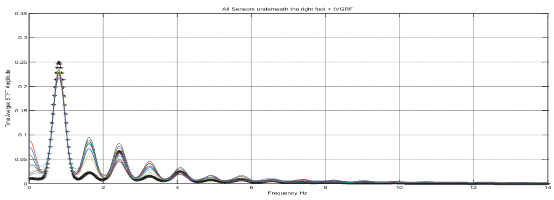

Fig.6: Amplitude of time averaged STFT amplitude for all 8 sensor underneath right foot for a normal subject. The strikes corresponds to frequency tVGRF.

Again, the main major frequency content is at a lower values. That's why WT is better used to tell more about the principal frequency components of gait VGRF. The first three peaks revealed a 94 % power from the whole spectrum. That's why they are considered.

Table. I demonstrate the first three fundamental peaks that cover the most power of the frequency content. The first peak is in the range of ~ 0.5658 – 1.1182 Hz for almost 98 % of subjects. It is approximately 2.0276 Hz for the second peak on average and 2.7108 Hz for the third peak.

TABLE I. FREQUENCY CONTENT

	Normal (mean ± std)		*Parkinson (mean ± std)*	
Peak #	***Frequency range (Hz)***	***Peak value***	***Frequency range(Hz)***	***Peak value***
1st	0.92 ± 0.07	~0.22	0.89 ± 0.11	~0.24
2nd	1.83 ± 0.13	~0.098	2.15 ± 0.73	~0.073
3rd	2.95 ± 0.9	~0.059	2.57 ± 0.72	~0.045

C. Differecence between Normal and parkinson

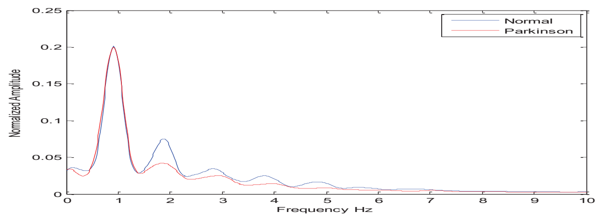

Fig.7: Amplitude of time averaged short time Fourier transform - Gaussian

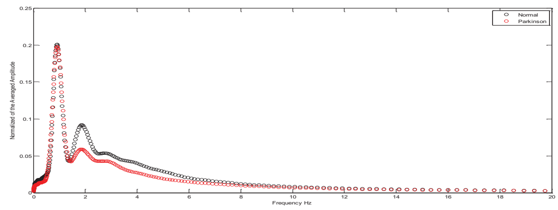

Fig.8: Amplitude of time averaged wavelet transform vs Frequency – Lognorm

The first peak covers the most important part of active part of the VGRF signal as shown in Fig.7. One research demonstrate at a cutoff frequency of 3Hz leads to imperfect separation of both active and impact peaks [13]. While it is not that such frequencies don't exist in Parkinson subjects as frequency tends to increase (i.e. the more they correspond to the impact peak), but the peaks and the valleys tend to be much more muted. That justifies the damping oscillations of the decomposed impact curve [13], on other words the result of spectral decomposition of VGRF. The lack of such peaks can be related to the stiffness or rigidity of limbs and consequently the postural instability that most Parkinson subjects will encounter. Nonetheless, Parkinson's restricted step length reduces their walking speed and number of steps per minute [14]. In order to discriminate the low frequencies (that usually forms the identity / main characteristics of the signal) between normal and Parkinson more clearly, the wavelet transform is used to provide higher temporal resolution for high frequencies as shown in Fig.8.

As a result, the first peak suggest the active curve and the second is due to impact peak. The latter is at higher amplitude in normal subjects. Table.I shows a coefficient of variation in the second peak of 7.04 % among normal subjects while it is 33.89 % among Parkinson. This is also confirmed by the boxplot shown in Fig.9 for 18 normal and 18 randomly selected Parkinson. The medians do differ with a 95% confidence since the notches in the box plot do not overlap.

From what is mentioned, this prove the difference that appears in the frequency oscillations at higher frequencies between normal and Parkinson gait. This totally agrees with the results of [11] that emphasize higher power at higher frequency in normal gait compared to a Parkinson gait. The latter shows smaller peak heights in the VGRF at heel contact and toe-off phases of gait. Such an impact peak becomes harder to be observed at advanced stages of disorder where steps of Parkinson characterized by shuffling steps.

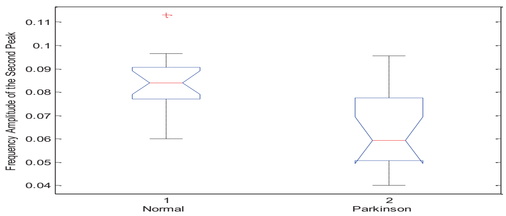

Fig.9: Amplitude of the second frequency component for both Normal and Parkinson subjects

V. Future Work

The frequency in VGRF is highly related to the walking speed of subject [16]. In addition, synochronsquezzing could be useful in differentiating normal gaits to those associated to a cognitive task. Higher frequencies corresponding to 5-15 Hz must be analyzed and their formation must be understood. Not to add, their variation could be useful to build a classifier between normal and diseased case. They could be also an indication for falling. Those are to be investigated more in a further studies by using those techniques in comparison to their alternative empirical mode decomposition.

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