Fourier analysis of tibia acceleration in subjects with knee osteoarthritis: preliminary results

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Abstract

Knee osteoarthritis (OA) can lead to pain and disability in the older population. Excessive loading has been suggested to contribute to the development of knee OA. In this paper, we investigated the frequency components in accelerometric signals of the tibia in the anterior-posterior (AP), medial-lateral (ML) and vertical (Vert) directions during walking between people with severe knee OA and asymptomatic control group. The Fast Fourier Transform (FFT) was applied to the tibia accelerations in the three directions for 3 trials. The FFT decomposes the acceleration signal into a series of signals at different frequencies to investigate the frequency components (harmonics) in a signal where the power spectrum graphs provide quantitative information of signal composition. The frequency components in the AP and ML accelerations appear at similar frequencies for both healthy and knee OA subjects. Frequency components in the vertical acceleration, however, were found to have larger frequency components in the higher frequencies (>5Hz) for the knee OA subjects compared to the healthy group. The differences in power peaks corresponding to higher harmonics (>6Hz) between the groups may indicate instability and altered attenuation of the impact during walking for people with knee OA. Further work, should be directed on investigating the position of occurrence of the high frequency during the gait cycle and understanding the biomechanical mechanism responsible for these differences.

1. INTRODUCTION

Knee osteoarthritis (OA) is one of the leading causes of disability among people aged 65 years and over in Australia [1]. Knee OA represents a major cause of disability and pain in the older population [2]. Knee OA is a multifactorial condition, with excessive joint loading being one of the main risk factors [3, 4]. Knee OA is believed to result from degenerative changes in the joint cartilage by deleteriously increase in joint loading [5].

Accelerometry has been proposed as a useful gait analysis technique to evaluate walking patterns and gait stability in different populations including people with knee OA [6-9]. Studies have used an accelerometer to investigate the load attenuation capacity of the tibia in medial-lateral, anterior-posterior and vertical directions in degenerative knees [7, 9, 10]. The results of these studies, however, were inconclusive partly due to the method used to measure tibial acceleration. A recent study found differences between symptomatic knee OA and healthy knees in the acceleration of the tibia in the medial-lateral and anterior-posterior directions [7]. This suggests that accelerometric parameters can be used to discriminate people with knee OA.

In this paper, we investigated the frequency components in an accelerometric signal of the resultant accelerations of the tibia in the anterior-posterior (AP), medial-lateral (ML) and vertical (Vert) directions during walking of subjects with severe knee OA (prior to undergoing knee replacement surgery). The Fast Fourier Transform (FFT) was used to decompose the acceleration signals into frequency components and ROC analysis was performed to determine the frequency components that had the highest dissimilarity between healthy and OA subjects. Our objective was to investigate if the frequency component of accelerometric data during walking can differentiate between people with knee OA and asymptomatic controls. The results have the potential to be used as follow up parameters for patients who are undergoing knee replacement surgery. As such, the accelerometric data will enable to identify improvement in the gait patterns following the surgery and therefore to assess patient recovery from surgery.

2. KNEE GAIT DATA

A. Subjects

The gait data consisted of two groups: symptomatic and age-gender matched asymptomatic control groups. The symptomatic group consisted of 4 patients (1 male and 3 females) with end-stage knee osteoarthritis one week prior to undergoing knee replacement surgery (with mean age,

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mass and height of 71.2 years ± 5.8, 82.9kg ± 6.8 and 167.6 cm ± 10.6 respectively). The asymptomatic control group consisted of 4 subjects (1 male and 3 females, with mean age, mass and height of 69.7 years ± 6.1, 70.7kg ± 14.3 and 161.1 cm ± 10.3 respectively).

B. Accelerometer Features

Linear accelerations of the tibia were measured along three orthogonal axes (vertical, anterior-posterior and medial-lateral) using a low mass tri-axial accelerometer. The accelerometer had a range of ±6g and was mounted on the skin over the tibia tuberocity with the vertical sensing axis aligned to the longitudinal axis of the tibia. The accelerometer was secured to the subjects using hypoallergenic waterproof tape (Leukoflex).

Kinematic parameters of tibia displacement in the three planes during level walking were recorded using a three-dimensional motion analysis system (Vicon MX3, Oxford Metrics, Oxford, UK) with 10 cameras (100Hz). The vertical ground reaction force was captured using two force plates (Kistler type 9865B, Winterthur, Switzerland and AMTI, OR6, USA) (1000 Hz) and was used to define the stance phase. Infrared retro reflective markers (14mm) were attached to anatomical locations on the lower extremity and Vicon Plug-in Gait (Oxford Metrics) biomechanical modelling software was used to process and output kinematic parameters. From the kinematic analyses three dimensional inclination of the tibia were obtained for use in the acceleration signal analysis to calculate the resultant acceleration of the tibia in three directions: vertical, anterior-posterior and medial-lateral.

Subjects were instructed to walk barefoot at a self-selected speed on a 10m walkway and three walking trials for the affected limb of the symptomatic group and the corresponding limb of the control group were used for the data analysis.

C. The Fourier Transform and Power Spectra

The discrete Fourier transform X(k) is used to represent a digital signal x(n) as a sum of fundamental functions (harmonics). This can be written in general form as:

$$X(k) = \sum_{1}^{N} x(n) e^{-j\omega_k n} \qquad (1)$$

where the angular frequency is $\omega_k = \frac{2\pi k}{N}$ for $k, n = 1, 2, ..., N$. The expansion of this is written as:

$$x(n) = \frac{1}{2} a_0 + \frac{1}{N} \sum_{k}^{N-1} a_k \cos\left(\frac{2\pi nk}{N}\right) + j b_k \sin\left(\frac{2\pi nk}{N}\right) \qquad (2)$$

where a_0 is the constant bias (DC, zero frequency), a_k and b_k are Fourier coefficients for harmonics $k = 1, 2, ..., N$ and have the same units as the signal $x(n)$. Fourier transforms are thus a powerful signal processing technique for providing information on signal frequency content. Since Fourier coefficients are unique, they can be used to characterize a signal. The strength (contribution) of the k^{th} harmonic in a signal can be measured by its power magnitude, defined as:

$$P(k) = a_k^2 + b_k^2 \qquad (3)$$

The power spectra of the signal x(n) is a graph of power magnitudes over all the harmonics in x(n) and can be used to quantitatively determine the major signal frequencies in gait activities.

3. Experimental Methodology

The Fast Fourier Transform (FFT) was applied to the resultant accelerations in the anterior-posterior (AP), medial-lateral (ML) and vertical (Vert) direction for 3 walking trials for each participant (4 subjects in each group). In this work, the 1000-point discrete FFT was computed using functions in Matlab's Signal Processing Toolbox, resulting in a 500 point spectral resolution (the other 500 are redundant) at a frequency range of 0-50Hz. The power spectra graphs were then calculated using equation (3). The maximum power peaks for integer frequencies (0,1,2,…,50Hz) were then found using a window of 5 samples across the frequency range.

The first 10 peaks (occurring at DC, 1Hz, 2Hz,…, 9Hz) were then extracted for all three acceleration directions for each subject resulting in 30 features for each subject. It was observed that harmonics after 9Hz were insignificant and were hence not included in the analysis. This coincides with previous findings that motion energy mainly lies in the low frequency ranges (0-5Hz) [11]. Subjects were then grouped into two classes. Those with knee OA were labelled with +1 while healthy control subjects were labelled -1. The mean, standard deviation and receiver operating curve (ROC) values were calculated for all 30 features (Table 1). The area of the ROC curve for single features was used to quantify the separability of the two groups based on the individual feature. ROC areas were numerically approximated using the trapezoidal rule where larger values implied better linear individual separability. In our case, higher ROC values meant that the particular feature (frequency harmonic) was more distinctive between the two classes.

Examples of frequency spectra for 2 subjects (asymptomatic control – CCB3 and knee OA – KRBJ1) are plotted for the vertical acceleration in Figure 1. The spectral peaks for the integer frequencies are plotted for 4 subjects for all three acceleration directions in Figure 2.

4. Experimental Results

A. FFT results

It can be seen from Figure 2 that a large DC bias was observed in the accelerations of all three planes which could be attributed to sensor bias. Movement activity was found to be approximately contained in the frequency ranges of (1-20Hz). It was observed that frequency components in the AP and ML accelerations appear at similar frequencies for both healthy and pathological subjects. Frequency components in the vertical acceleration were found to have larger frequency components in the higher frequencies (>5Hz) for the knee OA subjects compared to the healthy control group.

B. Statistical results

Table 1 depicts the results of the statistical analysis on the 30 features. It was found that the means of power peaks in the AP and ML directions to be relatively close to each other with comparatively large standard deviations, indicating that the acceleration trends in these planes were similar for both classes. This was further quantified by the single ROC feature values where the majority of values ranged between 0.56 and 0.75 for the AP direction and 0.5-0.75 for the ML direction indicating poor separation or minimal interclass difference. In the AP plane, the larger differences were found in higher order harmonics, e.g. 0.81 for the 8th harmonic and 0.94 for the 9th harmonic. In the ML plane, the major differences were found in the medium order harmonics as seen by the ROC values of 0.75 for 3rd harmonic and 0.94 for 4th harmonic.

The majority of the differences in frequency components were observed in the vertical acceleration. As seen in Table 1, the ROC values ranged from 0.5-1.0 across the frequencies. Larger differences between the two classes were observed in the medium harmonics e.g., 0.94 for 4th harmonic and higher harmonics e.g. 0.81 for 7th harmonic. The 8th harmonic of the vertical acceleration was found to completely characterize the difference between the classes with ROC=1.0. It was also observed that the mean magnitude of these peaks (power) was larger in the OA class compared to the asymptomatic class in these frequency regions.

5. Discussion

In this preliminary work, we have investigated the signal differences in the tibial acceleration in the AP, ML and Vert directions in a small sample group of patients with severe knee OA and asymptomatic controls. The harmonic signal represents the degree of smoothness and rhythm of the acceleration signal, therefore differences in the harmonic signal between the groups may indicate instability and altered attenuation of the impact during walking.

Previous study investigating peak acceleration of the tibia in people with knee OA reported differences in peak acceleration in the ML and AP direction [7] but no information on timing (frequency aspects) was investigated. In the current study, we found that differences in acceleration were evident in the higher order harmonics (8th and 9th) for the AP direction and (5th and 6th) in the ML direction. The power in these components was on average higher in the OA group than the control group, suggesting that motion in these directions were unstable.

These differences were more apparent in the vertical direction. The vertical frequency component depicts more apparent vibration during the gait cycle for the symptomatic group which may show instability and alteration in the transmission of shock to the knee during walking. At this juncture, it is unclear where the higher order frequencies occur in the gait cycle and hence further signal processing is required. The preliminary results however provide evidence that tibia accelerations provide useful information which can discriminate people with knee OA.

Our future work therefore is focused on localising the position of the highest frequency occurrence during the gait cycle and understanding the biomechanical mechanism responsible for these differences. Localization of the signal dissimilarities would allow us to focus on the gait events e.g. loading phase, stance or swing phases that are different in the OA group. This information would assist future analysis on accelerometer data as a sensor based solution to enable post-therapy monitoring and lead to new strategies for rehabilitation.

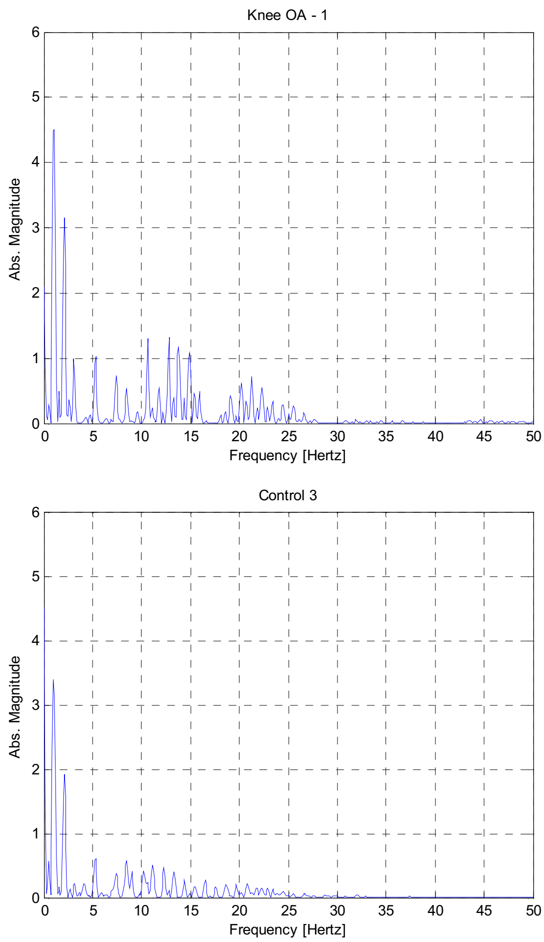
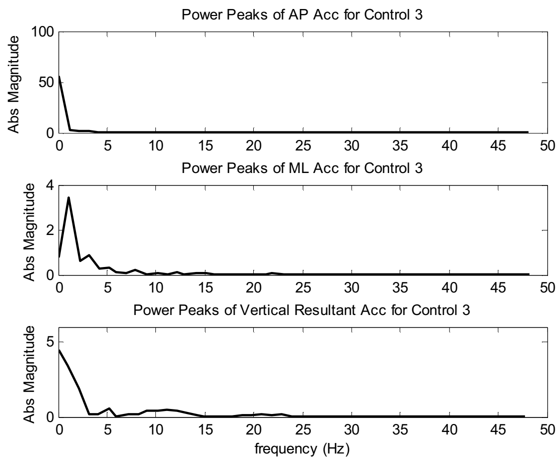


Figure 1: Representative sample of the vertical tibia acceleration Fourier spectra for knee OA subject and asymptomatic control subject.



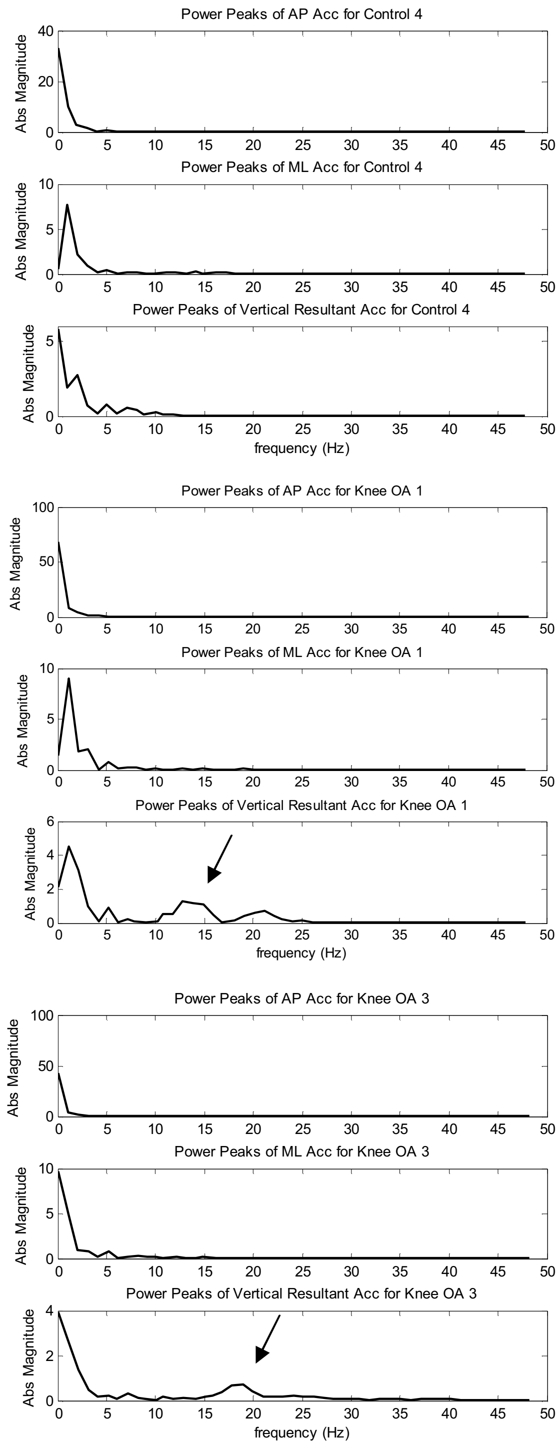


Figure 2: The spectral peaks for the integer frequencies for 4 subjects for the acceleration in the three directions (asymptomatic control 3 and 4 and knee OA 1 and 3). Note the presence of significant high frequency components in the vertical acceleration for Knee OA subjects compared to control subjects as indicated by the arrow.

TABLE 1: MEAN, STANDARD DEVIATION AND ROC CLASS SEPARABILITY FOR THE POWER SPECTRA OF AP, ML AND VERT ACCELERATION SIGNALS.

	AP-DC 1	AP-1 2	AP-2 3	AP-3 4	AP-4 5	AP-5 6	AP-6 7	AP-7 8	AP-8 9	AP-9 10
Mean Class 1	84.78	4.87	2.43	0.88	0.40	0.39	0.21	0.20	0.19	0.08
Mean Class -1	51.71	7.76	2.53	1.16	0.60	0.23	0.31	0.22	0.53	0.35
Std Class 1	36.97	2.16	1.62	0.44	0.49	0.40	0.13	0.20	0.22	0.05
Std Class -1	13.81	3.34	0.68	0.38	0.37	0.22	0.23	0.10	0.53	0.25
ROC Area	**0.81**	**0.75**	**0.63**	**0.75**	**0.75**	**0.56**	**0.63**	**0.63**	**0.81**	**0.94**

	ML-DC 11	ML-1 12	ML-2 13	ML-3 14	ML-4 15	ML-5 16	ML-6 17	ML-7 18	ML-8 19	ML-9 20
Mean Class 1	5.55	5.96	1.07	0.95	0.12	0.43	0.17	0.19	0.29	0.11
Mean Class -1	1.09	7.63	1.49	0.91	0.29	0.36	0.12	0.57	0.32	0.79
Std Class 1	4.22	2.11	0.55	0.78	0.09	0.37	0.19	0.04	0.17	0.04
Std Class -1	0.66	3.19	0.80	0.07	0.12	0.23	0.11	0.49	0.14	1.09
ROC Area	**0.94**	**0.63**	**0.69**	**0.75**	**0.94**	**0.63**	**0.63**	**0.63**	**0.50**	**0.63**

	V-DC 21	V-1 22	V-2 23	V-3 24	V-4 25	V-5 26	V-6 27	V-7 28	V-8 29	V-9 30
Mean Class 1	3.38	4.06	1.97	0.49	0.09	0.31	0.09	0.23	0.05	0.19
Mean Class -1	4.79	2.46	2.23	0.41	0.27	0.77	0.10	0.63	0.47	0.23
Std Class 1	1.31	1.10	0.82	0.39	0.07	0.42	0.10	0.14	0.04	0.21
Std Class -1	0.91	0.67	0.37	0.23	0.13	0.38	0.05	0.34	0.20	0.15
ROC Area	**0.81**	**0.94**	**0.75**	**0.50**	**0.94**	**0.81**	**0.69**	**0.81**	**1.00**	**0.63**

6. CONCLUSION

In this research, the harmonic content of tibia accelerometric signals was used to investigate differences in gait patterns of patients with knee OA. The study results have the potential to be used to assess patients' recovery by identifying the accelerometer pattern changes after knee replacement surgery. Further work, therefore, should be directed on investigating the occurrence of the high frequency during the gait cycle and understanding the biomechanical mechanism responsible for the differences between the two groups.

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