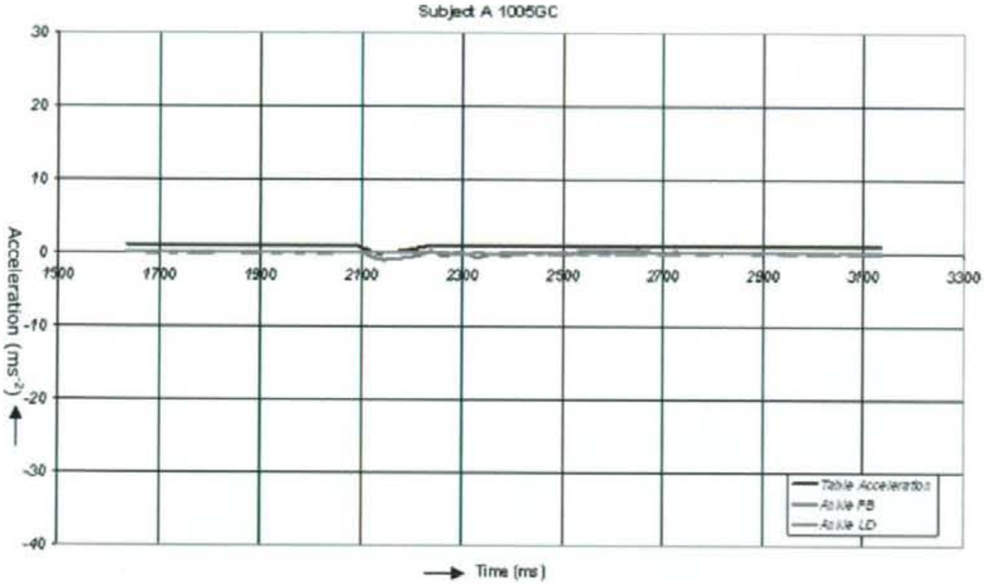

Fig. 5 Typical response, $1.0\,m/s^2$ (FB, front–back; UD, up–down)

Surrogate data were obtained by Fourier decomposition with the same amplitudes as the empirical data decomposition but with random phase components. This was achieved using the chaos data analyser [**23**]. Ten sets of surrogate data are generated for each person. The LLE is obtained for both the original and the surrogate data sets. It was found that the LLE for the surrogate data and that for the original data differ from each other by more than 58 per cent. This rejects the null hypothesis and hence the original data contain non-linear features.

3 RESULTS

The subject's response can generally be classified into three categories: first, no lifting of the soles; second, lifting of the soles; and third, stepping of the feet. Figures 5, 6, and 7 show the responses of a typical subject to acceleration magnitudes of $1.0\,m/s^2$, $4.0\,m/s^2$, and $4.8\,m/s^2$ respectively. Only the acceleration of the ankle in the front–back (dark grey curve) and up–down (light grey curve) directions are shown.

Table 1 shows the results of the LLE for ten subjects for ankle front–back acceleration and ankle pitch rate. Figure 8 shows the graph of Lyapunov exponent versus the acceleration magnitude.

The LLE quantifies the sensitivity of the system to initial conditions and gives a measure of predictability. This value decreases in various ranges with increase in the acceleration speed of the balance platform, indicating that the signal becomes less chaotic for normal subjects. The LLE corresponding to the ankle pitch rate and ankle front–back acceleration decreases with increase in the speed of the balancing platform.

4 DISCUSSION

From the raw data shown in Figs 2, 5, 6, and 7, it is easily apparent that at low acceleration magnitudes, only slight movement of the ankle occurs. During moderate acceleration magnitudes, there is greater ankle movement, and they occur over the entire period when the platform is moving. At high acceleration magnitudes, the ankle movement is very drastic, and it occurs only near the latter half of the platform movement.

Table 1 shows the values of LLE for different acceleration speeds of the balance platform. It was noted that the ankle front–back acceleration shows a clear decrease in the trend for the rise in the acceleration speeds.

The ankle sensor parameters (LLE) decrease as the acceleration magnitude becomes greater, and in-

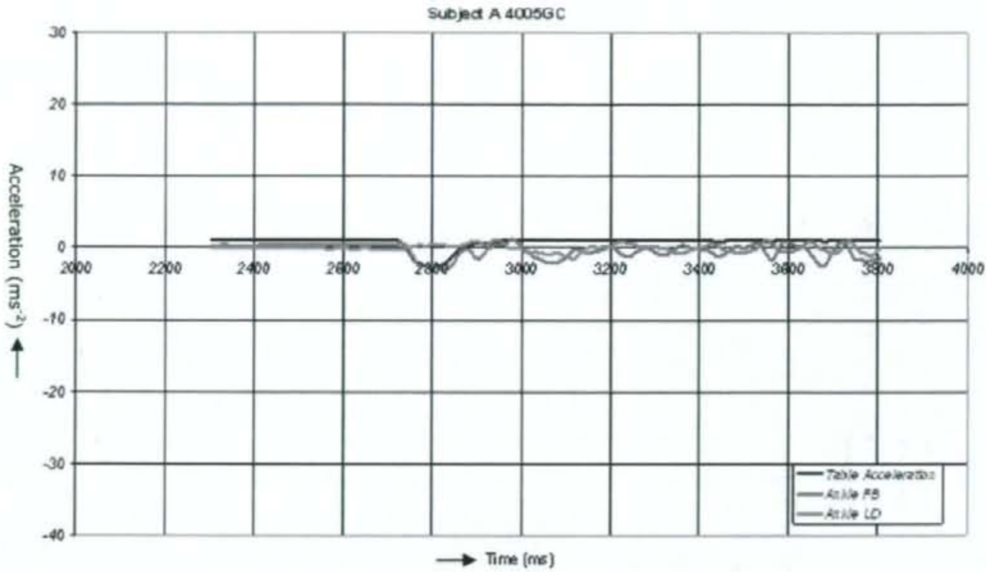

Fig. 6 Typical response, 4.0 m/s^2 (FB, front–back; UD, up–down)

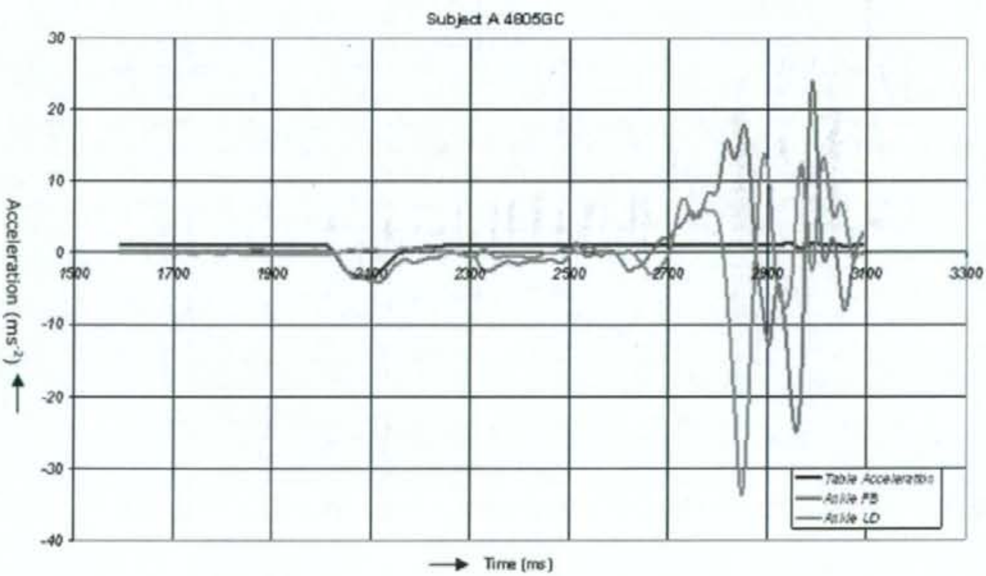

Fig. 7 Typical response, 4.8 m/s^2 (FB, front–back; UD, up–down)

creases slightly when the magnitude is less. This is because, when the magnitude is greater, the ankle moves more quickly in a particular rhythmic fashion. Therefore, these values will be smaller. However, when the acceleration magnitude is low, there is more random movement and hence these values will be slightly higher.

It is hypothesized that the drastic ankle movement (stepping) at high acceleration magnitudes have a hidden pattern and depend on the athletic ability of

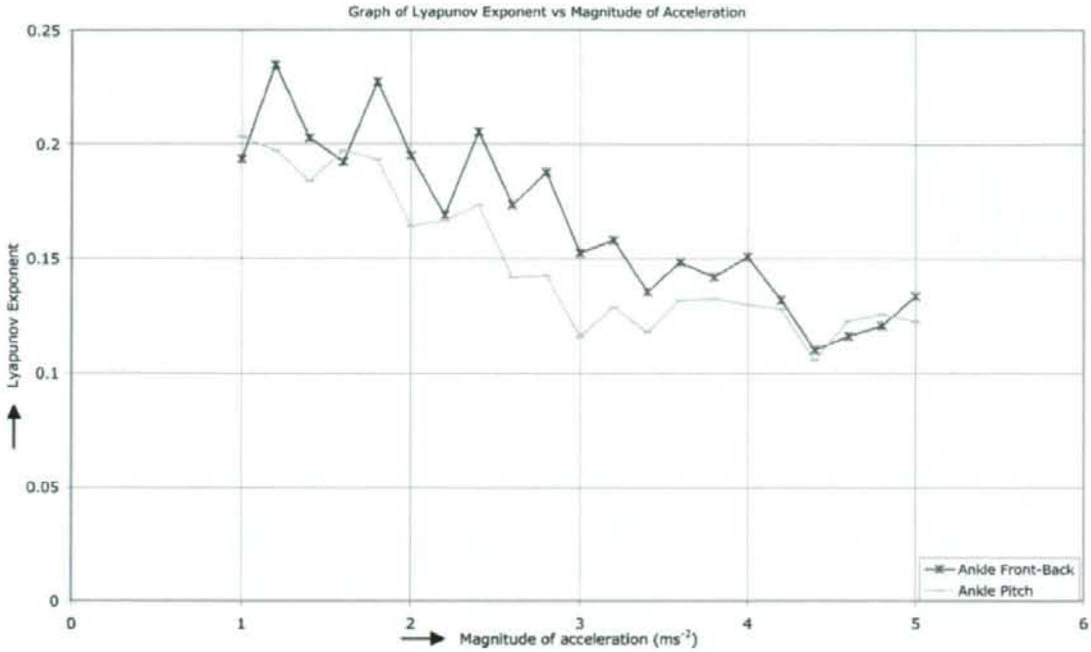

Fig. 8 Graph of the LLE versus the acceleration magnitude

the subject. The reason is that the subject will try to avoid falling as much as possible, exerting maximum effort. Hence the cadence of the stepping will be at the subject's 'favourite' pace.

For low acceleration magnitudes, fluctuations in ankle signals occur immediately after the acceleration. These are more random and hence result in higher LLE values.

The correlation dimension and the LLE studied on a time series of stabilograms showed distinct values in healthy subjects and Parkinsonian subjects [**24**]. Ohtaki *et al.* [**25**] have studied the Lyapunov exponents of young and elderly subjects, and of groups before after exercise intervention. Experimental results demonstrated that exercise intervention improved the local dynamic stability of walking. During

Table 1 Average LLE of ten subjects for ankle front–back acceleration and ankle pitch angular velocity

Acceleration magnitude (m/s^2)	Ankle front–back acceleration (m/s^2)	Ankle pitch angular velocity (degree/s)
1.0	0.1936 ± 0.001 26	0.2034 ± 0.001 012
1.2	0.2348 ± 0.001 144	0.1974 ± 0.000 684
1.4	0.2028 ± 0.002 954	0.184 ± 0.001 314
1.6	0.1924 ± 0.000 516	0.1972 ± 0.000 104
1.8	0.2274 ± 0.000 665	0.1934 ± 0.000 887
2.0	0.1952 ± 0.000 349	0.1642 ± 0.000 285
2.2	0.1692 ± 0.001 774	0.1668 ± 0.001 184
2.4	0.2056 ± 0.000 572	0.1734 ± 0.000 888
2.6	0.1736 ± 0.000 68	0.1418 ± 0.000 087 2
2.8	0.1878 ± 0.002 233	0.1426 ± 0.003 489
3.0	0.1526 ± 0.003 098	0.116 ± 0.002 078
3.2	0.1582 ± 0.002 545	0.1288 ± 0.000 283
3.4	0.1356 ± 0.004 318	0.118 ± 0.000 918
3.6	0.1484 ± 0.003 514	0.1318 ± 0.001 405
3.8	0.142 ± 0.004 707	0.1324 ± 0.001 019
4.0	0.1508 ± 0.003 807	0.13 ± 0.000 477
4.2	0.132 ± 0.002 526	0.128 ± 0.001 059
4.4	0.1104 ± 0.000 302	0.1058 ± 0.000 846
4.6	0.1162 ± 0.000 898	0.123 ± 0.000 454
4.8	0.1208 ± 0.000 166	0.1256 ± 0.000 254
5.0	0.1336 ± 0.001 243	0.1226 ± 0.000 299

quiet standing, the human body continually moves about in an erratic and possibly chaotic fashion. It has been shown that postural sway is indistinguishable from correlated noise and that it can be modelled as a system of bounded correlated random walks [**26**]. The results suggest that the postural control system incorporates both open-loop and closed-loop control mechanisms.

In this moving-platform experiment, at low acceleration magnitudes, the ankle movements are expected to be random, to maintain balance, and thus to confirm the findings. At high magnitudes, the ankle movements are less variable: first, before stepping occurs, they depend on the 'natural frequency' of the subject leaning back and forth in an oscillatory manner; second, after stepping occurs, they depend on the 'natural frequency' of the lower limbs and muscles, the subjects stepping at their favourite cadences. The present authors intend to continue this work by analysing the cadences of the same subjects on a treadmill to find their most efficient pace.

In this work, the responses of the ankle signal due to the movement of the acceleration platform have been analysed. The concept can be applied to analyse the quality of life of elderly subjects to perform their daily activities including the use of public transport.

5 SUMMARY

This work is focused on the application of a proper non-linear technique to analyse the complexity of postural signals. The surrogate data analysis applied to the data exhibited the deterministic and non-linear nature of the signals. The results of a force platform balance test suggest that the LLE for ankle front–back and ankle pitch rate decreases with increase in the balance platform acceleration. The proposed method was able to quantify the response of ankle signals to the external force (acceleration).

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