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APPLYING DATA CORRECTION TO STRAP MOUNTED ACCELEROMETERS

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INTRODUCTION

The tissue underlying skin mounted accelerometers introduces errors to the data they collect [1]. As a consequence various data correction attempts have been made to minimise the effect of local tissue-accelerometer vibration [1,2]. However, accelerometers are not always mounted directly onto the skin. It is often impractical to do so for studies that measure activities during day-to-day living where strap mounting may be a more common attachment method. Therefore an understanding of the response of strap mounted accelerometers is also necessary.

As the straps surround irregular shaped body segments strap mounted accelerometers may suffer from poor coupling when compared to skin mounted accelerometers, as well as additional vibration of the strap and pre-loading effects of tissue due to strap tension. This can be especially prevalent for straps around the waist, mounting accelerometers to measure motion at the spine. The aim of this study was to investigate whether the damped frequency (fd) and the logarithmic decrement (δ) of the local system (accelerometer, strap and local tissue) can be estimated so that the Smeathers' method of data correction [2] can be applied to strap mounted accelerometers at the lumbar spine.

METHODS

Six participants were fitted with tri-axial strap mounted accelerometers (GENEActiv, Action; 100 Hz) at the 4th lumbar vertebrae (L4). A variation on the nudge test [2] was performed whereby the strap was pulled down by approximately 2 cm and then released to provide an initial high acceleration disturbance whilst the participant was in a standing position. Three trials were completed and vertical acceleration data was imported into Excel to examine the accelerometer response.

From the decaying sinusoidal wave the fd and γ were calculated. The period between the first two successive negative peaks (A1 and A2) and their amplitudes from the steady state position (-1g) was determined. The fd was obtained from the reciprocal of the period and the δ calculated from the following equation [2]:

$$\delta = \log_e (A1/A2)$$

From the fd and γ the natural frequency and damping ratio of the local system was obtained using the equations previously reported by Smeathers [2].

RESULTS AND DISCUSSION

Following strap release the initial high acceleration response from the accelerometer showed a complex waveform which may represent multiple modes of vibration. Thereafter the response at lower acceleration levels showed the classic decaying oscillation of a simple mass, spring, damper system. The fd and δ were therefore obtained from the latter section of the accelerometer data (Figure 1).

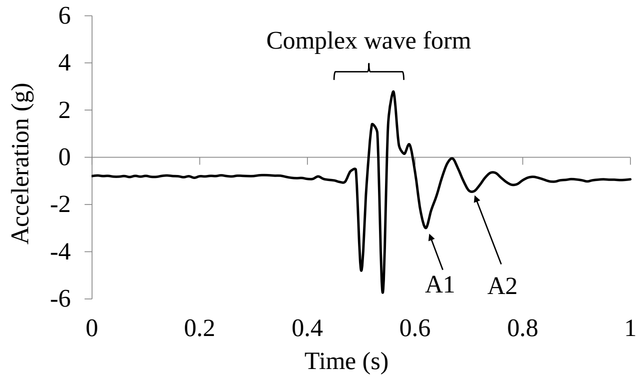

Figure 1: Typical accelerometer output showing the initial response to a high acceleration disturbance, the decaying oscillations and points of measurement.

The natural frequency ranged from 10.7 to 22.8 Hz, which was within previously reported ranges [2]. The damping ratio (0.18 to 0.30) at L4 was lower than reported previously (0.45) [2]. This is potentially due to the greater mass of the accelerometer (20g v 2.5g) and greater tissue mass due to strap mounting in the current study, as similar ranges (0.20 to 0.38) were reported for an accelerometer and mounting system of mass 25.4g attached to the skin at the 3rd lumbar vertebrae [1]. From the obtained natural frequency and the damping ratio an amplitude correction as function of frequency for the path from the spine to the accelerometer can be estimated from Smeathers' transmissibility equation derived from a linear spring, mass, damper model [2].

CONCLUSIONS

At lower levels of acceleration it is possible to measure the fd and δ for a strap mounted accelerometer system. These parameters can be used to estimate an amplitude correction as a function of frequency for each individual. Further work is needed to validate this transmissibility correction when appropriately applied to strap mounted accelerometer data of varying magnitudes and frequencies.

ACKNOWLEDGEMENTS

Funding was received from the University of Wolverhampton, Early Researcher Award Scheme.

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