THE EFFECT OF MECHANICAL SHOCK FREQUENCY AND AMPLITUDE ON SPINAL TRANSMISSION AND INTERNAL PRESSURE

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1. Introduction

When the human body is subject to vibration or impact, it demonstrates a dynamic response. The displacement of tissues and the forces transmitted by them alters as a function of time. A useful method to assess the potentially harmful effects is to measure the relative displacements and hence stresses of different regions of the body in response to vibration amplitude and frequency. Transmission of acceleration can be expressed in terms of a transfer function that defines the relative magnitude and phase relationship of the output acceleration in a particular region (for example, the spine) compared with the input acceleration (for example, at the seat). Knowledge of acceleration transfer functions provide insight into behaviour of the body sub-systems, and enables assessment of input acceleration levels and frequencies where a particular tissue is more likely to be damaged. A substantial body of knowledge has been reported concerning the transmission of vibration. However, little is known about the repeated impact environment and the dynamic response of individual body segments to vibration and impact. This paper reports some initial findings of the spinal and internal pressure responses to impact accelerations at the seat.

2. Methods

To determine health indices that could be used to develop a dose-effect model for minimizing the effects of repeated impact in army vehicles, a series of pilot experiments were conducted using the multi-axis ride simulator (MARS) at Fort Rucker, Alabama. These involved a series of seven 5.5 minute exposures to various shock frequencies, ranging from 2 to 11 Hz, and shock amplitudes, ranging from 0.5 to 3.0 g for each biodynamic axis (+x, +y and -z), with approximately 5 minutes rest between exposures. Each biodynamic axis was tested on a seperate day.

Ten male subjects between 20 and 40 years of age and within one standard deviation of the mean for height and weight, based on standard military data, were recruited from U.S. Army personnel assigned to Fort Rucker. All had experience with motion, either tactical ground vehicles (TGVs) or air transport. A solid metal seat with a bean-bag cushion and no backrest was securely mounted on the MARS. The seat was adjusted so the subject's feet rested comfortably on the MARS table with the knees and hips at approximately 90°.

Acceleration was measured at the seat and over the spinous process of lumbar (L2-4) and thoracic (T1-3) vertebrae using miniature Entran accelerometers (weight 0.3 gm; range of ± 10 g or ± 25 g). A skin to vertebrae inverse transfer function was generated using the method of Hinz et al. (1986) and applied to z- and y-axis spinal accelerations to correct measured accelerations for the elastic properties of the skin and subcutaneous tissues. The corrected accelerations represent motion of the vertebrae. Acceleration data were bandpass filtered at 0.5 to 60 Hz to eliminate baseline drift and high frequency noise.

Internal pressure was measured by a specially constructed rectal pressure probe, 50 cm in length and terminated in an Entran (model EPB-140W-5S) miniature pressure transducer (range ± 5 psi). The transducer and wiring were covered with 2 layers of heat shrink tubing 20 cm in length to provide a suitable degree of strength and flexibility. The probe was inserted by the subject to a depth of 15 centimeters beyond the anal sphincter. Internal pressure data were high pass filtered at 0.5 Hz to remove baseline fluctuations, and low pass filtered at 60 Hz to conform with seat acceleration data.

An acceleration peak detection and transmission ratio program was written to identify discrete shock events. The positive peak value of seat acceleration was first identified, followed by the acceleration minima (negative peak) occurring within a prescribed shock window of 250 msec. The program then analyzed spinal acceleration and internal pressure data to identify a corresponding positive and negative peak occurring within each shock window, and the time delay between the seat shock and peak response. Transmission ratios between seat acceleration and internal pressure were expressed as peak output pressure to acceleration input in units of $mmHg.s^2.m^{-1}$

3. Spinal Transmission of Shocks

Despite a range of responses between individuals, the shape of the mean transmission curves for seat to lumbar spine, seat to thoracic spine, and seat acceleration to internal pressure were remarkably similar for a given shock axis. There were, however, differences in magnitude and shape of acceleration responses between axes (x, y and z). The largest spinal responses were to -z-axis shocks. Maximal responses in the -z-axis were not, however, due to the initial shock input (the seat dropping out from below the subject), but rather to the secondary impact when the seat and subject were reunited. Acceleration responses to +y-axis shocks were higher in magnitude than to +x-axis shocks.

Spinal transmission curves for the x- and z- axis response to +x-axis shocks demonstrated non-linearity. At 3 g, there was a clear peak at 5 Hz in both the thoracic and lumbar spine and a smaller peak noticeable at 8 Hz. At 2 g and 1 g, however, the peak response shifted to 4 Hz and there was no second peak at 8 Hz. This pattern is clearly different from the dominant frequency of 1-2 Hz and diminishing response above 2 Hz suggested by ISO 2631 (1982) for sinusoidal vibration. Transmission of shock peaks was highest for 3 g shocks, followed by 2 g and then 1 g shocks. As the spine flexes in response to a shock, there are both x-and z-axis components due to curvature of the spine. The z-axis spinal response to x-axis shock inputs was higher than the x-axis response, likely due to rotation of vertebrae and whip-like motions.

In the y-axis, the shape of spinal acceleration response curves for lumbar and thoracic locations were similar for all shock amplitudes and for both y-and z-axis responses. The non-linearity of the response to x-axis shocks was not apparent for y-axis shocks. The dominant frequency was the lowest measured frequency (4 Hz for 3 g and 2 g; 2 Hz for 1 g). It is impossible to know whether the true dominant frequency is lower than these, since lower shock frequencies could not be measured. ISO 2631 suggests that the dominant frequency for sinusoidal vibration in the y-axis is 1-2 Hz. We cannot tell from this data whether the same is true for response to shocks. The z-axis spinal response to a y-axis shock was very small.

A non-linear response was apparent in the -z-axis, with highest responses recorded for higher amplitude shocks. The shape of spinal acceleration response curves for 3 g and 2 g shocks resembled that of the y-axis, with a dominant frequency of 4 Hz and a decreasing response at higher frequencies. Again, it is impossible to know whether a lower frequency shock will produce a higher response. For 1 g shocks, however, the dominant frequency was 4 Hz, not 2 Hz. This concurs with the ISO 2631 suggestion that the most sensitive frequencies for sinusoidal vibration are 4-8 Hz. The British Standards (BS 6841, 1987), which are meant to accommodate shocks, suggest that frequencies as high as 10 Hz are important in the z-axis. The DRI, designed for single large +z-axis impacts, suggests the dominant frequency is 11 Hz. Results from this study with -z-axis shocks, suggest the dominant frequency is lower than that in current standards.

The x-axis response at the spine to z-axis shocks at the seat was of similar magnitude and shape to the spinal z-axis response. As with x-axis shocks, the z-axis shocks caused a whip-like effect in the body resulting from forward flexion. At low frequencies, the x- and z-axis response at the lumbar spine averaged 2.5 times the seat acceleration. The upper thoracic spine showed a lesser amplification of seat input, approximately half that at the lumbar spine. Clearly, the body is a complex system that does not respond as a single mass to shock inputs.

When a subject impacts the seat following a -z-axis shock, the seat accelerometer registered a very high-frequency component (approximately 150 Hz). A corresponding high-frequency response (30-90 Hz) was transmitted to accelerometers at the spine. Data in the literature, however, suggests that the body damps these high frequency components (Fairley and Griffin, 1989). Our results showed higher spinal responses to 2 and 4 Hz shocks, compared with 11 Hz shocks. However, the much higher frequencies of 30-90 Hz were not being damped by the spine. It might be suggested that these high frequency components are skin movement. However, the vertebra-skin transfer function did not remove these. In addition, examination of internal pressure responses also showed similar high-frequency components. It may be that for a large enough impact the body responds as a non-linear system with transmission of high frequency components.

4. Internal Pressure Response to Shocks

The frequency dependence of internal pressure responses were remarkably similar to spinal acceleration responses, with a dominant peak commonly observed at 4 Hz (2 and 3 g) or 2 Hz (1 g) and a secondary peak at 8 Hz (3 g). The largest peak internal pressure response was measured in the z-axis (7.6 $mmHg.s^2.m^{-1}$ at 4 Hz). The lowest peak magnitudes were in response to y-axis shocks (0.86 $mmHg.s^2.m^{-1}$ at 4 Hz), however double peaks were often observed in the y-axis. The magnitude of internal pressure response to a 3 g impact in the z-axis varied from approximately 30 mmHg to 230 mmHg. By contrast, when subjects were asked to cough, they generated internal pressures of approximately 130 mmHg, laughing produced pressures of 80-90 mmHg, bearing down produced instantaneous pressures of 225 mmHg and sustained pressures of 130 mmHg. Some internal pressure magnitudes produced by 3 g shocks exceeded pressures that could be voluntarily produced by subjects.

5. Conclusions

The spinal acceleration and internal pressure responses to seat acceleration demonstrate clear non-linearities in the human response to mechanical shocks. The response curves are dependent on both shock magnitude and frequency.

These experiments highlight some limitations of current standards for vibration and shock. New standards for shocks should consider the non-linear response of the body and account for differences in the response to each axis (x, y and z) and different directions in each axes (+ and -). The shape of such a dose-response curve must also be magnitude dependent.

Further studies with shocks of lower frequency (<4 Hz for 3 and 2 g, and <2 Hz for 1 g) and higher magnitude (>3 g) are required to establish complete response curves with well defined maxima. It is also necessary to investigate the response to +z and -x shocks.

6. Note

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7. References

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