# The Evaluation Of Human Spinal Response To Vibration With Mechanical Shocks Of 0.5 To 4 G Amplitude

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## 1. INTRODUCTION

Mechanical shocks are a component of occupational whole-body vibration (WBV) exposure that may increase the risk of adverse health effects. While research has focused on the human response to WBV and to the very large impacts that occur during falls or vehicle crashes, very few studies have examined the response to low amplitude mechanical shocks (Robinson et al., 1993; Cameron et al., 1996). The ability to relate mechanical shock exposure to potential health effects depends, in part, on the ability to represent the human sensitivity to different characteristics of the motion.

Frequency weighting of the seat acceleration time history is utilized by the International Organization for Standardization (ISO 2631-1, 1997) to represent the human sensitivity to whole-body vibration. The Wk frequency weighting filter is used to represent the human response to vertical vibration. Payne (1992) and the Air Standardization Coordinating Committee (ASCC, 1982) proposed the use of the Dynamic Response Index (DRI) to represent the biodynamic response to vertical shocks of 1 to 20 g. The DRI is based on a second order linear system, initially defined by a natural frequency of 8.4 Hz and critical damping ratio of 0.224, but later revised to 11.9 Hz and 0.35 (Payne, 1992).

A series of experiments were designed to evaluate the human response to vertical mechanical shocks of varying amplitude, frequency content, and direction. This paper outlines some of the main characteristics of the human response to single mechanical shocks, and compares the measured biodynamic response to mechanical shocks with the estimation of human response afforded by the frequency weighting of ISO 2631-1 (1997) and the DRI (Payne, 1992).

## 2. METHODS

Experiments were conducted using the Multi-Axis Ride Simulator at the U.S. Army Aeromedical Research Laboratory, Fort Rucker, Alabama. Ten male volunteers were exposed to a variety of individual mechanical shocks ranging in amplitude from 0.5 to 4.0 g, in frequency from 2 to 20 Hz, in the +z (vertical) direction. Shock frequency was defined as the inverse of the time period of the biphasic shock waveform, where the shock waveform was presented as a damped sinusoid consisting of a single time period.

Acceleration was measured at the seat using three single axis Entran accelerometers (± 25 g) within a flexible epoxy seat pad that was securely taped to the seat cushion between the subject and the cushion. Acceleration at the spine was measured using Entran miniature accelerometers (± 25 g) attached to the skin over the thoracic and lumbar spinous process (at T3 and L4) by a small square of two-sided adhesive tape. Movement of the accelerometer and skin relative to the underlying bone was corrected using a transfer function derived from single perturbations of the accelerometer-skin system. Additional details of the experimental methods are contained in Cameron et al. (1996).

The Wk frequency weighting was applied to the measured z-axis seat acceleration using the transfer functions mathematically defined by Annex A of ISO 2631-1 (1997). The frequency response characteristics of the DRI were applied to the measured seat acceleration as a frequency weighting filter defined by the second order linear system parameters, fn=11.9 Hz and c = 0.35.

The ratio of peak acceleration response to peak acceleration input of the mechanical shock was calculated for the experimental data, the Wk filter, and the DRI model. This information was used to compare the predicted responses with the measured spinal response.

## 3. RESPONSE TO MECHANICAL SHOCKS

The main characteristics of the human response to individual vertical shocks with amplitudes greater than 1 g ($9.81\ m{\cdot}s^{-2}$) are illustrated in Figure 1. Results at the thoracic and lumbar vertebrae were similar. The acceleration data demonstrate two distinct events that influence the response at the lumbar spine: the initial mechanical shock and a secondary impact. The initial shock causes uncoupling of the seat and occupant, despite use of a seat belt, due to the phase lag in the response. When the uncoupling is reversed, there is a secondary impact. Both the initial shock and the secondary impact generate a significant response at the lumbar and thoracic vertebrae and have the potential to contribute to health effects.

While the human body tends to act as a low pass filter at low levels of vibration and shock, the body transmits higher frequency components as the magnitude of shocks increase above 2 g. This is visually evident in the response as higher frequency spikes superimposed on the lower frequency response (Figure 1).

In contrast, the Wk frequency weightings result in a response that more closely resembles the input acceleration at the seat than the response in the lumbar spine (Figure 1).

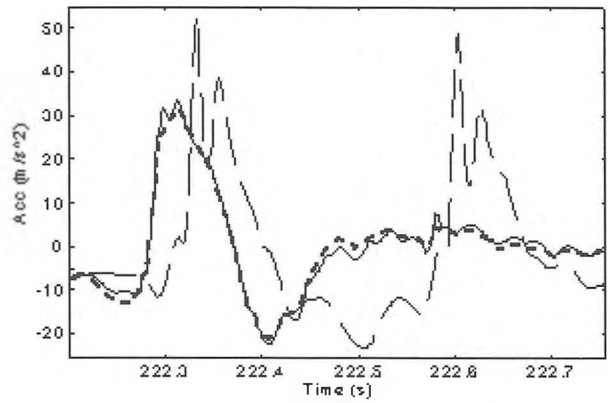


**Figure 1. Acceleration (m×s$^{-2}$) measured at the seat (solid) and lumbar spine (long dash) and Wk frequency weighted acceleration (short bold dash) for a +4 g, 5 Hz z-axis mechanical shock.**

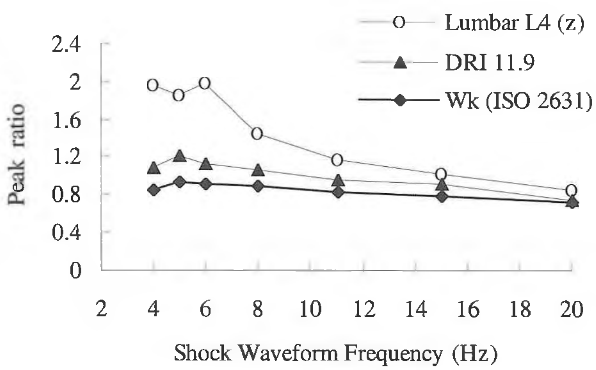


**Figure 2. Ratio of measured (Lumbar L4) and estimated (11.9 Hz DRI and Wk filter) peak response acceleration to peak acceleration at the seat for 4 g, z-axis mechanical shocks.**

Both the Wk filter and the DRI underestimate the measured response at all frequencies, and fail to predict the spinal response to the secondary impact (Figures 1 and 2).

These results suggest that the human spinal response in the z-axis represents a non-linear system, with the non-linear effects being dependent on both the amplitude and frequency (or period) of the shock waveform.

## 4. CONCLUSIONS

As shock amplitude and period increases in the z-axis, the human by transmission of higher frequency components and a response to both the initial shock and a secondary impact.

The nonlinear characteristics of the measured response to z-axis (vertical) mechanical shocks are not well represented by the frequency weighting filters of ISO 2631-1 (1997) or by linear system models such as the DRI. The DRI proposed by Payne (1992) has a natural frequency and damping coefficient that are not supported by the measured response data in these experiments. Both the DRI and the Wk filter underestimate the measured spinal response and fail to account for the response to the secondary impact.

An alternative approach is required if the human spinal response characteristics are to be adequately represented in evaluation of exposure to mechanical shock.

## 5. REFERENCES


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