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Novel methods to detect impacts within whole-body vibration time series data

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ABSTRACT
We present three candidate mathematical models for detecting impacts within time series accelerometer data in the context of whole-body vibration (WBV). In addition to WBV, data included recordings of erector spinae muscle activity and trunk posture collected during use of agricultural machines in a previous study. For each model, we evaluated associations between several mechanical and biomechanical variables at the time of predicted impact onset and the odds of subsequently observing a bilateral response of the erector spinae muscles. For all models, trunk posture at the time of impact onset was strongly associated with an observed bilateral muscle response; these associations were not observed when impacts were randomly assigned. Results provide a framework for describing the number and magnitudes of impacts that may help overcome ambiguities in current exposure metrics, such as the vibration dose value, and highlight the importance of considering posture in the evaluation of occupational WBV exposures.

Practitioner summary: Common metrics of exposure to whole-body vibration do not quantify the number or magnitudes of impacts within time series accelerometer data. Three candidate impact detection methods are presented and evaluated using real-world data collected during use of agricultural machines. Results highlight the importance of considering posture when evaluating vibration exposure.

Abbreviations: WBV: whole-body vibration; ISO: International Organisation for Standardization; ACGIH: American Conference of Governmental Industrial Hygienists; TLV: threshold limit value®; VDV: vibration dose value; RMS: root-mean-square; EMG: surface electromyography; US: United States; BMI: body mass index; IMU: inertial measurement unit; GLMM: generalized linear mixed model; OR: odds ratio; CI: confidence interval

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1. Introduction

Occupational exposure to whole-body vibration (WBV) is associated with low back problems (Bovenzi and Betta 1994; Wilder and Pope 1996; Bovenzi and Hulshof 1999; Lings and Leboeuf-Yde 2000; Bovenzi 2009; Burström, Nilsson, and Wahlström 2015; Kwaku Essien et al. 2018). Although the mechanisms through which WBV exposure leads to low back problems are not fully understood, high-risk occupational groups include operators of vehicles and machinery in transportation, construction, agriculture, forestry, mining, and military settings. The International Organisation for Standardisation (ISO) has developed standards for the measurement and evaluation of occupational WBV exposures, with aspects pertaining to WBV under general circumstances (i.e. ISO 2361-1) and when exposure includes repeated mechanical shocks or impacts (i.e. ISO 2631-5) (ISO 2010, 2018).

The ISO standards are the basis of occupational WBV exposure limits, such as the European Union's 2002/44/EC health and safety directive (European Union 2002), and consensus guidelines, such as the American Conference of Governmental Industrial Hygienists (ACGIH) Threshold Limit Value® (TLV) for WBV (ACGIH 2021). Several limitations of the ISO's WBV standards complicate the interpretation of measured vibration data and create challenges to evaluating potential health risks and implementing controls. Mathematically, a long-duration exposure to relatively steady-state vibration with several low-amplitude impacts could lead to the same 'vibration dose value'

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(VDV, from ISO 2631-1) or 'acceleration dose' (from ISO 2631-5) as a short-duration exposure with a vibration time history containing a few high-amplitude impacts (Rantaharju et al. 2015). Also, ISO 2631-1 suggests use of crest factor (the ratio of frequency-weighted peak to root-mean-square [RMS] accelerations) as one criterion for evaluating WBV exposure; when the crest factor exceeds 9 over the entire data record, both the frequency-weighted RMS value and VDV should be compared to recommended exposure limits. However, a large crest factor can be observed if a single high-amplitude impact occurs, which may confer a different amount of biomechanical loading than repeated impact events of lower peak amplitude (and for which the crest factor across the entire data record may not exceed 9). ISO 2631-5 partially compensates for this limitation by considering all accelerations that would cause spinal compression in the calculation of the acceleration dose.

However, the biomechanical responses to steady-state vibration are different than the biomechanical responses to impact events, and the biomechanical responses to both vibration and impact depend on seated posture (Pope et al. 1987; Zimmermann, Cook, and Goel 1993; Jack and Eger 2008; Desta, Saran, and Harsha 2011; Kumar, Saran, and Harsha 2017). Furthermore, the human responses to vibration and impact are normally evaluated in isolation, i.e. studies of the human response to vibration (without impacts) (Seroussi, Wilder, and Pope 1989; Zimmermann, Cook, and Goel 1993; Gant et al. 2012; Zhou and Griffin 2014) or studies of the human response to impact (without concurrent vibration) (Xia et al. 2008; Stenlund et al. 2018; Sivasankari and Balasubramanian 2020). As a result, the generalisability of these findings may be limited since vibration and impacts often occur together.

Although ISO 2631-1 and ISO 2631-5 provide useful analytic tools for evaluating occupational WBV exposures, neither standard (i) enumerates the number or magnitudes of impact events in the recorded data, (ii) addresses the role of paraspinal muscle activity to maintain spinal stability in vibration and impact environments, (iii) considers operator posture with respect to exposure evaluation (Annex B of ISO 2631-5 provides some informational content regarding posture, but posture *per se* is not factored into the calculation of acceleration dose), or (iv) considers how the human response to vibration and impact may adapt dynamically to changes in acceleration levels. Addressing these limitations may have important implications for understanding the mechanisms through which WBV exposure leads to low back problems and for informing the development of engineering or administrative control strategies.

In this paper, we present and evaluate three candidate computational models for detecting impacts in the context of concurrent vibration. Data for the analyses were obtained in the field among agricultural machinery operators in Midwest region of the United States and included: acceleration at the seat/operator interface, posture of the trunk (obtained using inertial sensors), and erector spinae muscle activity (obtained using surface electromyography [EMG]) (Fethke et al. 2018; Fethke et al. 2020).

2. Methods

2.1. Study participants and source data

Participants ($n = 14$) were a subset of 518 farmers enrolled in a longitudinal study of physical risk factors and musculoskeletal pain among agricultural workers in the US Midwest (Fethke et al. 2015). The current study included measurement datasets from 14 participants while operating 19 machines (of 112 total in the original dataset). Thirteen participants were male, and two were left hand dominant. Participants' mean age was 62.5 years (range: 51.6–79.3 years), and mean body mass index (BMI) was 28.1 $kg{\cdot}m^{-2}$ (range: 17.7–44.3 $kg{\cdot}m^{-2}$). Machines considered in the current study were selected to encompass a range of WBV exposure levels and included seven combines, 10 tractors, one skid loader, and one forklift (with pneumatic tires). From each measurement dataset, data were extracted from the full recording during a single period over which, based on visual inspection and on notes made by research assistants, the participant was on the machine seat and the machine was running continuously. Across the 19 machines, the duration of the extracted analysis period averaged 31.2 minutes (range: 5.9–61.6 minutes). Mean measured WBV exposure levels, based on ISO 2631-1, included: total frequency-weighted RMS acceleration = 0.52 $m{\cdot}s^{-2}$ (range: 0.20–1.18 $m{\cdot}s^{-2}$); crest factor (z-axis) = 9.40 (range: 5.16–16.19), and VDV (z-axis only) = 4.90 $m\,s^{-1.75}$ (range: 2.02–12.05 $m{\cdot}s^{-1.75}$).

2.1.1. Muscle activity measurement

Continuous EMG recordings were obtained bilaterally from the thoracic erector spinae, with electrodes positioned 5 cm lateral to the 9th thoracic spinous process (McGill 1991). Two EMG data logger systems were used: the Delsys Myomonitor IV® with DE2.3 differential electrodes (Delsys Inc., Boston, MA, USA; electrode bandwidth 20–450 Hz and gain = 1000) and the Biometrics DataLog with SX230 differential electrodes (Biometrics Ltd., Gwent, UK; electrode bandwidth 20–460 Hz and gain = 1000). For both systems, a reference electrode was positioned over the clavicle on the non-dominant side and the raw EMG signals were digitised at 1000 Hz.

2.1.2. Trunk posture measurement

Continuous recordings of trunk posture were made using two inertial measurement units (IMU; Series SXT, Nexgen Ergonomics, Inc., Quebec, Canada). As reported previously (Fethke et al. 2018), one IMU was positioned over sternal notch and a second IMU was positioned over the posterior pelvis at the L5/S1 level. For each IMU separately, raw triaxial accelerometer (± 6 g) and gyroscope (± 2000° s^{-1}) data were sampled at 20 Hz and then combined using a first-order complementary weighting algorithm to estimate pitch angle (Schall et al. 2015; Chen, Schall, and Fethke 2018; Chen, Schall, and Fethke 2020). In the current study, 'trunk inclination angle' refers to the pitch angle of the IMU located over the sternal notch (with respect to gravity) and 'back flexion' refers to the difference in the pitch angles of the two IMUs. For each participant, the IMUs were calibrated using the manufacturer's recommended procedures (i.e. an 'I-pose' reference posture).

2.1.3. Vibration measurement

Unweighted acceleration at the seat/operator interface was recorded using a semi-rigid, triaxial seat pad accelerometer (model 356B41, PCB Piezotronics, Depew, NY), in accordance with ISO 2631-1 (ISO. 2010). In addition, unweighted z-axis (vertical) acceleration was recorded using a single axis accelerometer (model 353B33, PCB Piezotronics, Depew, NY) attached to the seat baseplate, machine floor, or frame as near as possible to midline of the seat base. All acceleration signals were digitised at 1280 Hz using a data recorder (DA-20, Rion Co., Ltd., Tokyo, Japan).

2.2. Conceptual framework

Before describing the candidate impact detection models, our overall conceptual framework and criteria for what constitutes an impact must be defined. For simplicity, each candidate model considers only acceleration along the z-axis (vertical) recorded at the seat/operator interface. We made three overarching assumptions:

1. Any acceleration signature characterised as a simple sinusoid will contain no impacts.
2. An acceleration signature characterised by a slow and steady increase in RMS amplitude will contain no impacts.
3. The system is casual, i.e. whether an acceleration signature at any one point in time constitutes an impact depends solely only the characteristics of the preceding acceleration signature.

Our approach assumes that the erector spinae EMG signals (at any point in time) are in response to maintaining spinal stability. We also assume that the EMG response to an impact (if observed) is stimulated by the onset of the impact event (Xia et al. 2008). Finally, we assume there exists a 'dynamic threshold' that, when exceeded during an impact event, will elicit an identifiable erector spinae muscle response. The dynamic threshold concept considers that increasing amplitude of steady-state vibration will cause a concurrent increase in EMG activation (Blüthner, Seidel, and Hinz 2002) and, thus, decrease the likelihood of observing an increase in EMG activation following a non-steady-state perturbation. For example, the erector spinae EMG signals of a person sitting quietly in the absence of vibration may increase briefly following a small perturbation of the seat; however, the same perturbation may not elicit a similar EMG response if the seat is also vibrating.

2.3. Defining the response to an impact event

Broadly, we define the response to an impact event as an increase in EMG signal amplitude during the time following the impact onset in comparison to the time preceding the impact onset. The raw EMG signals were high-pass filtered (6th order Butterworth, zero-phase) with a 70 Hz cut-off frequency to maximally attenuate motion and electrocardiogram artefacts. The greater than typical high-pass filter cut-off frequency may also result in EMG signal information more strongly correlated with muscle force production (Potvin and Brown 2004). The high-pass filtered EMG signals were then converted to instantaneous RMS amplitude using a 25 ms sliding window (Xia et al. 2008).

From the time series of instantaneous RMS EMG amplitudes, the lowest values observed (for the right and left erector spinae separately) over the entire analysis period were defined as the 'baseline noise.' All instantaneous RMS EMG amplitudes were then adjusted for baseline noise using quadratic subtraction (Thorn et al. 2007). Then, given an instant in time for which the onset of an impact is identified (methods described in Section 2.4, below), two windows of EMG data were extracted: (i) a 1.0 s window beginning 1.05 s preceding the impact and ending 0.05 s preceding the impact (i.e. the pre-impact window), and (ii) a 0.6 s window beginning 0.1 s following the impact and ending 0.7 s following the impact (i.e. the post-impact window). The pre-impact window timing and duration were selected to minimise risk of attributing EMG activity caused by active postural adjustments (e.g. participants shifting their position on the machine seat) to the pre-impact EMG response to the vibration environment. The post-impact window timing and duration were based on prior research suggesting that (i) the onset of the erector spinae response occurs no earlier than 50 ms following an unexpected stimulus (Cavanagh and Komi 1979; Seidel, Bluethner, and Hinz 1986; Seroussi, Wilder, and Pope 1989; Granata, Slota, and Bennett 2004) and (ii) the peak of the erector spinae response to an unexpected vertical impact generally occurs within ~300 ms (Xia et al. 2008). We doubled the window duration to account for uncertainty arising from the complexity of our real-world source data.

Two criteria were used to declare the presence or absence of an erector spinae response. First, at least one RMS EMG amplitude within the post-impact window must exceed a value equal to the mean plus three standard deviations of the RMS EMG amplitudes within the pre-impact window. Second, the integral of the RMS EMG amplitudes within the post-impact window must exceed 0.6 times the integral of the RMS EMG amplitudes within the pre-impact window. The 0.6 multiplier compensates for the difference in the duration of the pre-impact and post-impact windows. The muscle response was analysed as a dichotomous variable, coded as '1' if each criterion was met for both the right and left erector spinae muscles and coded as '0' otherwise.

2.4. Impact detection methods

Participants operated machinery as they normally would, so some shifting of body position on the seat was inevitable duration data collection. To ensure that our impact detection methods would not interpret acceleration related to shifts in body position as candidate impact events, the instantaneous magnitude of the seat acceleration was limited to 1.10 times the maximum (or minimum) acceleration at the machine floor during the preceding 0.5 s. Each of the 19 measurement datasets was analysed with each impact detection method, described below.

2.4.1. The 'thump' method

The thump method is inspired by VDV, but with a few key differences. First, the acceleration data are not frequency-weighted. Second, rather than computing a single VDV over the entire recording, the acceleration time history is parsed into non-overlapping epochs of 0.1 s in duration. For each epoch, the mean acceleration amplitude is subtracted from each acceleration value, the result is raised to the 4th power, and then a 'thump value' is obtained by integrating the mean-adjusted 4th power acceleration values over the epoch duration (resulting in units of $m^{4}s^{-7}$) (Equation 1). Third, in contrast to the VDV, we do not take the 4th root of the integral result. The thump value reflects the range of acceleration experienced within each 0.1-s epoch. The determination of what is/is not an impact is then made by treating the dynamic threshold as a first-order system with the thump value as the input (Equation 2). If the thump value exceeds the threshold, then we assume an impact has occurred (Equation 3).

$$thump(t) = \int_{t_0}^{t_0+0.1} \left(a(t) - \frac{1}{0.1}\left(\int_{t_0}^{t_0+0.1} a(t)dt \right) \right)^4 dt \quad (1)$$

$$thresh(t) = thump(t) - \left[\sqrt{2}\left(thump(t) \otimes \left(e^{-0.921t} * u(t-0.1) * u(t-5)\right)\right) \right] \quad (2)$$

$$impact\left(t\right) = \left\{ \begin{array}{l} 1 \;\; if \;\; thresh(t) > 0 \\ 0 \;\; if \;\; thresh(t) \le 0 \end{array} \right\} \quad (3)$$

The constants used in Equation 2 assume that (i) the system impulse response decays to 1% of its original magnitude after five seconds before immediately going to zero and (ii) identical impacts (without concurrent vibration) must be separated in time by at least 0.4 s to be considered independent events with respect to the erector spinae muscle response. If *impact(t)* = 1 (i.e. an impact occurred), then the instant of impact onset is identified as the instant of maximum jerk within the epoch.

2.4.2. The 'womp' method

The womp method identifies regions of sustained jerk substantially exceeding that observed during the preceding time. Conceptually, this approach considers that (i) erector spinae muscle activity will increase in response to sustained increases in vibration magnitude (i.e. an acclimation process) but that (ii) brief lulls in vibration magnitude will not cause an immediate desensitisation to the previous vibration stimulus. To minimise error in the calculation of jerk resulting from numerical differentiation of acceleration, the acceleration data were first frequency-weighted in accordance with ISO 2631-1 (ISO. 2010). At each sample (i.e. delta-t = 1/1280s), we define a new dynamic threshold by calculating the RMS value plus two standard deviations of the jerk from the preceding 7.0 s and 1.0 s and taking maximum of these two values. If the jerk exceeds this dynamic threshold for at least 10 ms, then we assume an impact has occurred. The impact onset is then assigned as the instant of maximum jerk over the duration for which the dynamic threshold is exceeded. Finally, we calculate a 'womp value' as the integral of the jerk over the duration of the impact (resulting in units of m·s^{-2}).

2.4.3. The 'wiggle' method

The wiggle method identifies impacts similar to the manner in which the ISO 2631-5 standard identifies local acceleration maxima (ISO. 2018). Essentially, each bounce (or 'wiggle') of the operator is separated into negative velocity (i.e. gravitational force-dominant) and positive velocity (i.e. seat force-dominant) phases, and then analyses are focussed on the positive velocity phase. The acceleration data are frequency-weighted in accordance to ISO 2631-1, numerically integrated to velocity, and then AC-coupled using a zero phase high pass filter. We then (i) calculate a 'wiggle value' as the integral of the 4th power acceleration between each local velocity minimum and maximum (Equation 4), and (ii) apply a modified version of Equation 2 to adjust the dynamic threshold, compensating for the elimination of regions with net negative acceleration by doubling the coefficient (i.e. from $\sqrt{2}$ to $2\sqrt{2}$) (Equation 5).

$$wiggle(t) = \int_{t@\min(t)}^{t@\max(t)} a(t)^4 dt \tag{4}$$

$$thresh(t) = wiggle(t) - \left[2\sqrt{2}\left(wiggle(t)\otimes \left(e^{-0.921t} * u\left(t - t@min(t) + t@min(t-1)\right) * u(t-5)\right)\right)\right] \tag{5}$$

$$impact(t) = \left\{ \begin{array}{l} 1 \;\; if \;\; thresh(t) > 0 \\ 0 \;\; if \;\; thresh(t) \leq 0 \end{array} \right\} \tag{6}$$

The impact onset is defined as the instant of the local velocity minimum. Since velocity is 180° out of phase with jerk in pure sinusoidal vibration, the instant of a local velocity minimum would correspond to the instant of a local jerk maximum. In comparison to the thump method, the wiggle method considers only positive acceleration (i.e. between local velocity minima and maxima) and the duration of each analysed region of positive acceleration varies (vs. a fixed 0.1 s epoch duration).

2.4.4. The 'random' method

In the absence of gold standard data needed to make a definitive judgement about what is or is not an impact, the random method was used as a baseline with which to evaluate the ability of the thump, womp, and wiggle methods to identify impacts resulting in a bilateral erector spinae response. As the name implies, the random method assigns 'impacts' at random times throughout each measurement dataset. For each measurement dataset separately, the number of impacts assigned was the greater of (i) the maximum number of impacts identified using the thump, womp, and wiggle methods or (ii) the duration of the extracted analysis period (in seconds) divided by 30 (i.e. one impact for every 30 seconds). The locations (in time) of each impact were then randomly seeded. The onset of each impact was then assigned to the instant of maximum jerk within a window spanning 0.05 s before to 0.10 s after the randomly seeded impact location.

2.5. Mechanical and biomechanical variables

For each impact identified using each method (i.e. thump, womp, wiggle, and random), we calculated a set of mechanical and biomechanical variables to consider as predictors of an erector spinae muscle response. Mechanical variables included: *maximum jerk* from 0.05 s prior to the impact onset to 0.10 s following the impact onset (absolute value, in m·s^{-3}); *VDV*, calculated using the frequency-weighted acceleration (z-axis only) from the beginning of data record to the instant of impact onset (in m·s$^{-1.75}$); the *time since previous impact* (in s); and *peak acceleration* (unweighted) in the z-axis direction from 0.05 s prior to the impact onset to 0.10 s following the impact onset (absolute value, in m·s^{-2}). Biomechanical variables included: *trunk inclination angle* (in °) at the sample nearest in time to the instant of impact onset and *back flexion angle* (in °) at the sample nearest in time to the instant of impact onset.

2.6. Statistical analyses

Our primary objective was to estimate the strength of association between the mechanical and biomechanical variables and the odds of observing a bilateral erector spinae response. To accomplish this objective, generalised linear mixed models (GLMM) were

Table 1. Number and magnitudes of impacts detected and the number of impacts eliciting a bilateral erector spinae response, by impact detection method.

	Random ($n = 3242$)		Thump ($n = 2107$)		Womp ($n = 2659$)		Wiggle ($n = 3757$)	
Peak Accel. ($m \cdot s^{-2}$)	*N* (%)	Resp. (%)[a]	*N* (%)	Resp. (%)[a]	*N* (%)	Resp. (%)[a]	*N* (%)	Resp. (%)[a]
Total		513 (15.8)		337 (16.0)		426 (16.0)		648 (17.2)
<0.31	881 (27.2)	132 (15.0)	102 (4.8)	18 (17.6)	190 (7.1)	35 (18.4)	234 (6.2)	36 (15.4)
0.31 to < 0.61	840 (25.9)	139 (16.5)	253 (12.0)	43 (17.0)	521 (19.6)	76 (14.6)	525 (14.0)	95 (18.1)
0.61 to < 1.23	951 (29.3)	153 (16.1)	759 (36.0)	130 (17.1)	920 (34.6)	151 (16.4)	1118 (29.8)	199 (17.8)
1.23 to < 2.45	394 (12.2)	59 (15.0)	601 (28.5)	75 (12.5)	817 (30.7)	121 (14.8)	1086 (28.9)	177 (16.3)
2.45 to < 4.90	155 (4.8)	28 (18.1)	287 (13.6)	52 (18.1)	175 (6.6)	38 (21.7)	569 (15.1)	108 (19.0)
4.90 to < 9.81	21 (0.6)	2 (9.5)	90 (4.3)	16 (17.8)	30 (1.1)	4 (13.3)	194 (5.2)	27 (13.9)
≥ 9.81	0 (0.0)	0 (0.0)	15 (0.7)	3 (20.0)	6 (0.2)	1 (16.7)	31 (0.8)	6 (19.4)

[a]Number (%) of impacts eliciting a bilateral erector spinae muscle response.

constructed with the muscle response analysed as a dichotomous variable (as described in Section 2.3). A separate GLMM was constructed for each impact detection method. Our expectation was that the mechanical and biomechanical variables would be more strongly associated with a muscle response for impact events identified using the thump, womp, and wiggle methods than for impact events identified using the random method.

The mechanical and biomechanical variables were each categorised using quartiles prior to entering them into the GLMMs as fixed effects. Quartiles were defined separately for each impact detection method, and for each variable were based on the distribution among impact events for which a bilateral muscle response was observed. This approach ensured an equal number of muscle response events within each quartile, thus minimising cell size discrepancies that could reduce precision of the resulting odds ratios. Age and BMI were entered into the GLMMs as continuous (fixed) covariates. Each GLMM also included a random intercept term to account for clustering of observations at the individual level.

In addition to the primary analyses, a second set of GLMMs was constructed to estimate the strength of association between variables unique to each impact detection method (i.e. thump value, womp value, and wiggle value) and the odds of observing a bilateral erector spinae response. Each initial GLMM included as fixed effects (i) the unique impact detection variable (in quartiles), (ii) each mechanical and biomechanical variable (in quartiles, as above), and (iii) age and BMI (as continuous variables, as above), as well as a random intercept term (as above). A modified backward elimination procedure was then used to estimate adjusted associations between the impact detection variables and the odds of observing a muscle response. Specifically, the least significant covariate (i.e. all other variables except the unique impact detection variable) with $p > 0.2$ was removed first, followed by the next least significant covariate with $p > 0.2$, and so on until all covariates with $p > 0.2$ were removed. If the removal of any covariate resulted in a change of >10% in the association between the unique impact detection variable and muscle response, then that covariate was returned to the model before the next covariate was removed. Therefore, each final model included the unique impact detection variable (which was forced into the model and not subject to removal) and any covariate that was associated with the muscle response (at $p \leq 0.2$) or confounded the association between the unique impact detection variable and the muscle response, regardless of its *p*-value. All statistical procedures were performed in SAS (version 9.4, SAS Institute Inc., Cary, NC, USA).

3. Results

The total number of impacts identified (across all measurement datasets) was 2107 using the thump method, 2659 using the womp method, and 3757 using the wiggle method (Table 1). The proportion of impacts for which a bilateral erector spinae response was observed was only marginally greater for the thump (16.0%), womp (16.0%), and wiggle methods (17.2%) in comparison to the random method (15.8%). However, examining the number of impacts according to the peak z-axis acceleration at the seat confirmed that the thump, womp, and wiggle methods identified events of greater acceleration magnitude than the random method. Specifically, the peak acceleration was $\geqq 0.61\ m \cdot s^{-2}$ for ~47% of the impact events identified using the random method but for ~73–83% of the impact events identified using the thump, womp, and wiggle methods.

Consistent with the above, the median peak z-axis acceleration at the seat for impacts identified using the thump, womp, and wiggle methods was roughly double that for the random method (Table 2). The median maximum jerk was marginally greater for impacts identified using the thump and wiggle methods than for the

Table 2. Median and interquartile [25th percentile, 75th percentile] ranges[a] for each mechanical and biomechanical variable across all impacts, by impact detection method.

		Random		Thump		Womp		Wiggle	
Variable	Response	median	[25th, 75th]	median	[25th, 75th]	median	[25th, 75th]	median	[25th, 75th]
Maximum jerk ($m \cdot s^{-3}$)	Yes	18.8	[9.1, 38.1]	20.4	[7.8, 45.9]	56.1	[37.5, 79.1]	27.1	[17.1, 51.1]
	No	18.1	[9.1, 38.7]	22.7	[7.6, 48.4]	56.4	[37.3, 84.1]	29.8	[17.6, 57.1]
VDV ($m \cdot s^{-1.75}$)	Yes	3.3	[2.5, 4.0]	3.2	[2.4, 4.0]	3.3	[2.5, 3.9]	3.1	[2.1, 4.0]
	No	3.3	[2.5, 4.0]	3.3	[2.6, 4.0]	3.2	[2.4, 3.9]	3.2	[2.1, 4.5]
Time since previous (s)	Yes	5.9	[2.2, 13.7]	7.9	[3.6, 17.9]	5.0	[1.9, 12.7]	6.5	[3.4, 12.1]
	No	5.9	[2.3, 13.9]	8.4	[3.7, 17.6]	5.0	[1.9, 11.3]	6.1	[2.9, 11.1]
Peak seat accel. ($m \cdot s^{-2}$)	Yes	0.6	[0.3, 1.0]	1.1	[0.7, 2.0]	1.0	[0.6, 1.6]	1.2	[0.7, 2.1]
	No	0.6	[0.3, 1.0]	1.2	[0.8, 2.0]	1.0	[0.6, 1.6]	1.2	[0.7, 2.2]
Trunk inclination angle (°)[b]	Yes	−13.0	[−19.5, −7.6]	−16.0	[−24.0, −8.7]	−14.3	[-19.3, −7.8]	−16.3	[-26.2, −7.7]
	No	−13.5	[−20.0, −7.0]	−14.0	[−21.0, −7.2]	−12.8	[-18.4, −7.5]	−14.9	[-22.4, −6.9]
Back flexion angle (°)	Yes	29.8	[17.5, 37.7]	32.0	[18.0, 42.0]	30.1	[16.2, 37.8]	33.0	[21.0, 40.0]
	No	30.7	[17.1, 38.5]	30.0	[15.0, 39.0]	29.1	[16.1, 36.1]	32.0	[19.0, 40.0]
Thump value ($m^4 \cdot s^{-7}$)[c]	Yes		n/a	0.03	[0.007, 0.17]		n/a		n/a
	No		n/a	0.03	[0.009, 0.16]		n/a		n/a
Womp value ($m \cdot s^{-2}$)[c]	Yes		n/a		n/a	0.12	[0.06, 0.24]		n/a
	No		n/a		n/a	0.13	[0.06, 0.30]		n/a
Wiggle value ($m^4 \cdot s^{-7}$)[c]	Yes		n/a		n/a		n/a	0.16	[0.04, 1.30]
	No		n/a		n/a			0.17	[0.04, 1.72]

[a]Quartiles for statistical analysis derived from the median and interquartile range among impacts with response = yes.
[b]Negative trunk inclination angle reflects trunk extension with respect to gravity, as with a slightly reclined posture.
[c]Thump value from Equation 1; womp value = integral of jerk over impact duration; wiggle value from Equation 4.

random method. However, the median maximum jerk for impacts identified using the womp method was more than triple that for the random method, which reflects the womp method's reliance on jerk. The median, 25th percentile, and 75th percentile values for trunk inclination angle were negative for all impact detection methods, which reflects a predominantly reclined posture among the participants during machine operation. On the other hand, the median, 25th percentile, and 75th percentile values for back flexion angle were positive, reflecting the loss of lumbar curvature.

Odds ratios (OR) describing associations between the mechanical and biomechanical variables (as well as age and BMI) and a bilateral erector spinae response are provided in Table 3. For impacts identified using the random method, no variable was significantly associated with a muscle response. In contrast, increasing trunk inclination angle (i.e. from reclined to more upright) was significantly associated with a decreased odds of observing a bilateral erector spinae response for impacts identified using the thump (quartiles 3 vs. 1: OR = 0.56, 95% CI = 0.37–0.87; quartile 4 vs. 1: OR = 0.50, 95% CI = 0.31–0.81) and wiggle (quartile 3 vs.1: OR = 0.69, 95% CI = 0.51–0.93; quartile 4 vs. 1: OR = 0.68, 95% CI = 0.48–0.96) methods. The pattern was similar for the womp method, although associations were not statistically significant. The associations between back flexion angle and a bilateral muscle response generally mirrored those for trunk inclination angle, i.e. the ORs increased as back flexion angle increased (particularly for the thump method). In general, across the impact detection methods, no other consistent patterns were observed in the associations between the mechanical and biomechanical variables and the muscle response.

Adjusted associations between the unique impact detection variables and a bilateral muscle response were predominantly not statistically significant (Table 4). As the lone exception, impacts with a wiggle value in the 4th quartile were significantly more likely to elicit a response than impacts with a wiggle value in the 1st quartile (OR = 1.72, 95% CI = 1.18–2.51).

4. Discussion

In this study, we developed three candidate mathematical models for detecting impact events from complex, real-world acceleration data obtained during operation of agricultural machinery. Performance of each model was compared to performance of a 'random' model in which the timing of impact events was randomly allocated across the acceleration time series. Performance in this context was defined as associations between variables describing mechanical (e.g. peak acceleration) and biomechanical (e.g. posture) parameters during each impact event and the odds of observing a bilateral response of the erector spinae muscles.

Although the theoretical underpinnings and computational approaches of the three candidate models differ, the trunk inclination and back flexion angles at the time of impact onset were most strongly and consistently associated with the odds of observing a muscle response. Specifically, muscle responses were less likely as the trunk inclination angle changed from

Table 3. Multivariable associations between mechanical, biomechanical, and personal (age and body mass index) variables during an impact event and the odds of observing a bilateral erector spinae response (OR = odds ratio).

	Random			Thump			Womp			Wiggle		
Variable	OR	[95% CI]	p^a	OR	[95% CI]	p^a	OR	[95% CI]	p^a	OR	[95% CI]	p^a
Maximum jerk			0.49			0.38			0.19			0.53
Quartile 2 (vs. 1)[b]	0.90	[0.68–1.19]		1.32	[0.92–1.89]		0.98	[0.69–1.41]		1.02	[0.78–1.35]	
Quartile 3 (vs. 1)	1.11	[0.83–1.49]		1.01	[0.71–1.45]		1.02	[0.69–1.50]		0.90	[0.66–1.22]	
Quartile 4 (vs. 1)	1.08	[0.77–1.53]		1.13	[0.78–1.63]		0.70	[0.45–1.09]		0.82	[0.59–1.16]	
VDV			0.56			0.46			0.20			0.24
Quartile 2 (vs. 1)	1.00	[0.74–1.38]		1.06	[0.70–1.60]		1.28	[0.87–1.86]		1.00	[0.75–1.33]	
Quartile 3 (vs. 1)	0.84	[0.61–1.16]		0.89	[0.58–1.36]		1.51	[1.03–2.19]		1.04	[0.75–1.43]	
Quartile 4 (vs. 1)	0.90	[0.64–1.25]		0.78	[0.50–1.22]		1.29	[0.88–1.88]		0.76	[0.53–1.09]	
Time since previous impact			0.99			0.70			0.87			0.20
Quartile 2 (vs. 1)	0.95	[0.72–1.25]		1.00	[0.70–1.41]		0.98	[0.72–1.33]		1.24	[0.97–1.59]	
Quartile 3 (vs. 1)	0.98	[0.74–1.29]		0.86	[0.60–1.21]		0.89	[0.66–1.21]		1.15	[0.90–1.47]	
Quartile 4 (vs. 1)	0.98	[0.73–1.32]		0.86	[0.60–1.23]		0.98	[0.70–1.35]		1.28	[1.00–1.65]	
Peak z-axis acceleration at the seat			0.84			0.26			0.80			0.69
Quartile 2 (vs. 1)	1.13	[0.86–1.48]		1.25	[0.87–1.80]		1.17	[0.84–1.63]		1.04	[0.80–1.37]	
Quartile 3 (vs. 1)	1.09	[0.82–1.46]		0.90	[0.62–1.30]		1.13	[0.80–1.60]		1.14	[0.84–1.54]	
Quartile 4 (vs. 1)	1.06	[0.77–1.46]		0.95	[0.62–1.46]		1.07	[0.72–1.60]		1.25	[0.86–1.81]	
Trunk inclination angle			0.18			0.03			0.04			0.06
Quartile 2 (vs. 1)	1.08	[0.79–1.49]		0.74	[0.49–1.12]		1.14	[0.81–1.60]		0.84	[0.63–1.13]	
Quartile 3 (vs. 1)	1.42	[1.00–2.01]		0.56	[0.37–0.87]		0.72	[0.49–1.04]		0.69	[0.51–0.93]	
Quartile 4 (vs. 1)	1.12	[0.78–1.62]		0.50	[0.31–0.81]		0.74	[0.49–1.11]		0.68	[0.48–0.96]	
Back flexion angle			0.57			0.02			0.25			0.30
Quartile 2 (vs. 1)	1.10	[0.78–1.56]		1.20	[0.77–1.88]		0.84	[0.56–1.24]		1.10	[0.82–1.46]	
Quartile 3 (vs. 1)	0.92	[0.63–1.34]		1.57	[1.00–2.48]		1.11	[0.71–1.73]		1.27	[0.95–1.69]	
Quartile 4 (vs. 1)	0.89	[0.60–1.31]		2.18	[1.32–3.62]		1.21	[0.76–1.93]		1.26	[0.92–1.71]	
Age	1.00	[0.97–1.02]	0.94	1.00	[0.97–1.03]	0.89	1.02	[0.99–1.05]	0.11	1.02	[1.00–1.04]	0.04
Body mass index	1.01	[0.98–1.04]	0.60	1.01	[0.96–1.06]	0.62	0.99	[0.94–1.05]	0.72	0.98	[0.95–1.01]	0.15

[a] *p*-values reflect the results of Type III tests of fixed effects from the generalised linear mixed model procedure.
[b] Quartile ranges were calculated separately for each impact detection method; see median [IQR] values from Table 2.

Table 4. Adjusted associations between unique impact detection method metrics and the odds of observing a bilateral erector spinae response (OR = odds ratio).

	Thump value[a]		Womp value[b]		Wiggle value[c]	
	OR	[95% CI]	OR	[95% CI]	OR	[95% CI]
Quartile 2 (vs. 1)[d]	0.68	[0.46–1.00]	1.15	[0.85–1.55]	1.20	[0.91–1.57]
Quartile 3 (vs. 1)	0.80	[0.52–1.23]	0.98	[0.71–1.34]	1.19	[0.87–1.61]
Quartile 4 (vs. 1)	1.06	[0.63–1.80]	0.80	[0.56–1.10]	1.72	[1.18–2.51]

[a] Adjusted for peak z-axis seat acceleration ($p = 0.14$), trunk inclination angle ($p < 0.01$), and back flexion angle ($p < 0.01$).
[b] Adjusted for trunk inclination angle ($p = 0.15$).
[c] Adjusted for maximum jerk ($p = 0.12$), VDV ($p = 0.15$), time since previous impact ($p = 0.09$), and age ($p = 0.11$).
[d] See median [IQR] values from Table 2 for quartile ranges.

a reclined position to a position closer in alignment with the gravity vector. Conversely, muscle responses were more likely as the back flexion angle became more positive, i.e. as the angle between the anterior torso and posterior pelvis increased. From a biomechanical perspective, an operator sitting with the spine aligned with the z-axis would be less likely to have a large muscle response to impacts directed along this axis because muscle forces needed to counter the moment generated by the weight of the torso and head are minimised.

To our knowledge, the muscle response to impact events has never been evaluated using data generated outside a laboratory environment. Therefore, there are few (if any) previous studies to which our results can be compared directly. Others, however, have reported impact detection methods. For example, Allen, Taunton, and Allen (2008) reported a method of detecting impacts from acceleration signals recorded from a tri-axial accelerometer affixed to the deck of a rigid-hull inflatable boat. Two trials were conducted at-sea (traveling 15–20 knots), including periods of relative calm and with sea states of 2–3 (depending on trial). Their algorithm returned 2184 impacts during a 70-min trial and 2181 during a 90-min trial (or impact rates of about 24–31/min). Using the mean duration of the extracted analysis periods in this study (i.e. 31.2 min) as the denominator, our models returned impact rates from 68 to 120/min considering all impacts identified and from 11–21/min considering only impacts for which a bilateral muscle response was observed. In addition to environmental differences, differences in impact rates may be related in part to our use of dynamic thresholds for defining an impact compared to the constant threshold used in Allen, Taunton, and Allen (2008). Also, the peak acceleration exceeded 1 g for about 44% of the z-axis impacts detected in Allen, Taunton, and Allen (2008), and impacts up to 9 g were reported. In contrast, peak acceleration of impacts detected in this study rarely exceeded 1 g. Although Allen, Taunton, and Allen (2008) calculated VDV from frequency-weighted acceleration signals, the acceleration was not measured at the seat/operator interface and so their results are not comparable to the VDVs reported in this study.

Brammer, Roddan, and Morrison (2010) also reported a method for detecting impacts, which was based on statistical properties of the acceleration data. Two criteria were used to declare the presence of an impact within an acceleration time series of arbitrary duration: (i) the ratio of the 12th-order root mean acceleration to the RMS acceleration and (ii) an 'impulsiveness' metric. Each criterion was defined using the 97th percentile of the cumulative probability of acceleration amplitudes within the time series under the assumption of random vibration with zero average value. Brammer, Roddan, and Morrison (2010) demonstrated the utility of their method using simulated signals and limited field data (with time series durations of $\sim$20 seconds). However, in contrast to the current study, no physiologic data were available with which to explore the biological relevance of their method. Furthermore, Brammer, Roddan, and Morrison (2010) did not present a method for either counting the number of impacts or defining the temporal location of impacts within the acceleration time series.

4.1. Methodological considerations

The thump method identifies localised acceleration segments characterised by substantially larger acceleration ranges than during the preceding time. However, the model does not distinguish between negative and positive acceleration, and does not preserve temporal information about the acceleration profile. Similar to VDV, the same thump value may be obtained from a smooth increase in acceleration magnitude over the 0.1 s epoch and from a single, near-instantaneous high-amplitude shock. We elected to compute a thump value for non-overlapping epochs of 0.1 s in duration. Alternatively, the thump value could be generated as a continuous time series. The epoch-based approach is less computationally intensive than the continuous approach and ensures that each thump value is calculated from unique acceleration data. However, the epoch-based approach could truncate an impact signature in such a way that it does not exceed the threshold. On the other hand, calculating a continuous thump value could potentially smear an impact signature due to the convolution with the decay function. Future analyses could explore epoch-based approaches that incorporate different epoch durations and/or overlapping (e.g. half-window).

In contrast to the thump method, which relies on acceleration, the womp method relies on locating regions of sustained jerk that exceed what is considered a statistically predictable jerk threshold. The conclusion drawn from this is that when an unexpected impact occurs, the body reacts as though the jerk was expected to be maintained indefinitely, and subsequently the muscle response 'overshoots' with contraction forces greater than necessary to resist the moment induced by the impact. Such a response has been observed in previous studies examining the biomechanics of sudden loading events (Marras, Rangarajulu, and Lavender 1987; Mannion, Adams, and Dolan 2000; Grondin and Potvin 2009; Shahvarpour et al. 2014), and has timing characteristics consistent with a reflex response (Granata, Slota, and Bennett 2004).

The strength of association between the wiggle value and muscle response suggests that the responses are stimulated by positive accelerations and the macro-motion of the seat rather than the higher frequency components. This result offers empirical support for the methods used in ISO 2631-5. The wiggle model might also serve as a basis to analyse impacts within vibrating environments by allowing one to describe the 'wiggles' as being vibration-caused or impact-caused, or to parse a rapid change of acceleration into vibration and impact components.

4.2. Strengths and limitations

Strengths of the current study include examining individual impact events as unit of analyses rather than the machines or subjects themselves ($n > 2000$ impacts per method vs. $n = 19$ machines or $n = 14$ subjects); using real-world, complex data to evaluate our models; and the introduction of the dynamic threshold concept to the evaluation of the response to impacts within the context of concurrent vibration.

The current study represents an initial step towards overcoming certain limitations to evaluating occupational WBV exposure using the ISO standards. Specifically, vibration profiles with an identical VDV can be differentiated according to the number and distribution of peak accelerations of impacts within the recorded time series data. As an initial step, however, several important limitations must be considered. Foremost, our approach examines only vertical acceleration at the seat/operator interface, whereas in some environments fore-aft and lateral acceleration may be dominant and muscle responses to non-vertical perturbations of the trunk are dissimilar from those to vertical perturbations. For example, Xia et al. (2008) observed faster muscle response times and greater muscle response amplitudes on the side opposite a lateral impact, whereas response times and amplitudes were similar bilaterally when the impact was vertical. Extending our models to incorporate fore-aft, lateral, or total acceleration may yield different results. Furthermore, because the data were collected during actual agricultural production and not in controlled laboratory conditions, we cannot be certain that the observed muscle responses truly reflected biodynamic behaviour. Synchronisation of the EMG, IMU, and WBV data streams was based on the internal clocks of the instrumentation systems rather than an external trigger, which also could have introduced small errors. Importantly, the source data did not permit our models to consider sensory cues (outside of the experienced acceleration) that might allow participants to anticipate an impact event. Sensitivity analyses with modifications to the timing and/or duration of the pre- and post-impact windows could provide additional insights regarding this issue. We also acknowledge that every assumption and design choice made in developing our impact detection methods, while grounded in prior research, could be challenged, altered, or rendered irrelevant with future study.

Finally, the seats of all machines considered in the current study included a back rest, although some were designed to support only the lower back (e.g. in the skid loader, forklift, and smaller tractors) while others supported both the lower and upper back regions (e.g. the combines and larger tractors). However, our analyses could not consider the possible influences of either the design or operators' use of the back rest on the erector spinae response to impacts.

5. Conclusion

Ultimately, one goal of detecting the number and estimating the magnitude of impacts within WBV (acceleration) data is to allow increased specificity of WBV exposures evaluated using metrics from the ISO 2631-1 and 2631-5 standards. For example, with further development, results from an impact detection method presented in the current study could be used to scale or weight the VDV metric in such a way that eliminates the mathematical possibility of obtaining identical VDVs from acceleration profiles with disparate temporal characteristics. Although none of the methods presented performed appreciably 'better' than the others, an important observation is the strength of association between operator posture at the time of impact onset and the likelihood of observing a bilateral erector spinae muscle response. Muscle forces contribute substantially to spinal loads, and so the results strongly support consideration of posture in the evaluation of the potential health risks from occupational exposure to WBV. This could form the foundation for evaluation of fatigue of muscle, hard tissue and connective tissue as both and/or either, passive and/or active materials, based on the characteristics of the load cycles, analogous to fatigue analysis of standard engineering materials or failure of mechanisms due to component timing or impedance mismatching.

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