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Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration —

Part 5: Method for evaluation of vibration containing multiple shocks

Vibrations et chocs mécaniques — Évaluation de l'exposition des individus à des vibrations globales du corps —

Partie 5: Méthode d'évaluation des vibrations contenant des chocs répétés

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Foreword

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The procedures used to develop this document and those intended for its further maintenance are described in the ISO/IEC Directives, Part 1. In particular the different approval criteria needed for the different types of ISO documents should be noted. This document was drafted in accordance with the editorial rules of the ISO/IEC Directives, Part 2 (see www.iso.org/directives).

Attention is drawn to the possibility that some of the elements of this document may be the subject of patent rights. ISO shall not be held responsible for identifying any or all such patent rights. Details of any patent rights identified during the development of the document will be in the Introduction and/or on the ISO list of patent declarations received (see www.iso.org/patents).

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For an explanation on the voluntary nature of standards, the meaning of ISO specific terms and expressions related to conformity assessment, as well as information about ISO's adherence to the World Trade Organization (WTO) principles in the Technical Barriers to Trade (TBT) see the following URL: www.iso.org/iso/foreword.html.

This document was prepared by Technical Committee ISO/TC 108, *Mechanical vibration, shock and condition monitoring*, Subcommittee SC 4, *Human exposure to mechanical vibration and shock*.

Any feedback or questions on this document should be directed to the user's national standards body. A complete listing of these bodies can be found at www.iso.org/members.html.

This second edition cancels and replaces the first edition (ISO 2631-5:2004), which has been technically revised. The main changes compared to the previous edition are an improved description of the physiological response function for the exposure and improved guidance on the associated risk.

A list of all the parts in the ISO 2631 series can be found on the ISO website.

Introduction

The purpose of this document is to define a method of quantifying whole-body vibration containing multiple shocks in relation to human health in the seated posture. In biodynamics, the term “shock” is used to describe a wide range of short-time, high-magnitude exposures. It covers the range of severity starting at mild shocks resulting only in annoyance and brief discomfort up to magnitudes of shock sufficient to cause pain, injury or substantial physiological distress.

The methods described in this document can be appropriate for assessing the risk of chronic injury from exposure to repeated shock as can be experienced in military, commercial or recreational off-road vehicles, including agricultural vehicles, heavy plant equipment and high-speed marine craft. The methods are not intended to assess the probability of acute damage from a single impact.

The assessment methods described are based on the predicted biomechanical response of the bony vertebral endplate (hard tissue) in an individual who is in good physical condition with no evidence of spinal pathology. However, the risk assessment methods and related models described in this document have not yet been systematically epidemiologically validated. The methods provide nevertheless a quantitative description of the exposure, which is necessary to assess relative differences between exposures, e.g. the effects of some protective measures and different exposure conditions.

This document solely addresses lumbar spine response on the basis of studies indicating that the lumbar spine can be adversely affected by exposures to whole-body vibration[6][7][8][9][10][11][38][39][47][48][54][55] which also contain multiple shocks. Other adverse health effects of exposure to repeated shock, such as damage to parts of the body other than the lumbar spine, or types of short or long term health effects other than damage to the vertebral end plates, are not specifically considered by this document. Such end plate damage often cannot be differentiated by damages caused by other exposures (heavy lifting) and diseases.

This document considers only the effects of compressive loads from multiple shocks. To this end, a seat-to-lumbar spine transfer function of the measured acceleration has been developed for a default posture, body height and lumbar spine level. Another method to describe the spinal response is given in Annex A, which is valid only for a limited range of acceleration magnitudes but includes the effect of different postures, body heights and lumbar spine levels.

A standardized approach to the prediction of injury for non-vertical or combined axes shocks is complicated by the range of different postures and body restraint systems that can be employed in different vehicles and the limitations of current capabilities for predicting injury from non-vertical shock. Shocks involving horizontal, rotational or multi-axial motion are known to occur in practice and can present a significant risk of injury.

The risk of injury in the lumbar spine depends on an exposure dose, which is a combination of an exposure quantity and a duration. A manifest injury can take several years to develop. Due to the complexity of the measurement of multiple shocks, it is at the moment not possible to measure the exposure of the lifetime dose directly. Instead, the exposure is measured in representative situations and the dose is extrapolated from this measurement to a recorded exposure duration in the past or an anticipated exposure duration in the future. To monitor constantly the lifetime dose at a workplace, alternative measurement equipment will need to be developed, e.g. dosemeters.

Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration —

Part 5:
Method for evaluation of vibration containing multiple shocks

1 Scope

This document addresses human exposure to multiple mechanical shocks, and it formulates requirements for the measurement of multiple shocks. The results of these measurements are then analyzed to provide information for the assessment of the risk of adverse health effects to the vertebral end-plates of the lumbar spine for seated individuals due to compression. Other injuries could develop even when there is no injury to the end plate.

NOTE 1 Multiple mechanical shocks are shocks of different magnitude and shape that occur frequently at regular and irregular intervals during the measurement period.

NOTE 2 As proposed in the annexes, the assessment of the current injury risk is based on measured representative exposures in combination with the individual exposure history. Prospective risks can be assessed by anticipated exposure durations. Manufacturers of measurement equipment are encouraged to develop a possibility for an on-site evaluation of the exposure.

Two exposure regimes are distinguished in this document: one for severe conditions and one for less severe conditions.

NOTE 3 Clause 4 contains the delineation of the two regimes.

This document is applicable for unweighted vertical accelerations that have peak values up to 137,3 m/s^2 (14 g) measured at the seat-occupant interface beneath the ischial tuberosities over a 0,01 Hz to 80 Hz measurement bandwidth.

NOTE 4 The measurement bandwith is defined in 5.1.

Caution is necessary when applying the method to severe exposures, particularly since peak accelerations of 137,3 m/s^2 (14 g) are close to the physical limit that a spine can tolerate.

2 Normative references

The following documents are referred to in the text in such a way that some or all of their content constitutes requirements of this document. For dated references, only the edition cited applies. For undated references, the latest edition of the referenced document (including any amendments) applies.

ISO 2041, *Mechanical vibration, shock and condition monitoring — Vocabulary*

ISO 2631-1:1997, *Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration — Part 1: General requirements*

ISO 5805, *Mechanical vibration and shock — Human exposure — Vocabulary*

ISO 10326-1, *Mechanical vibration — Laboratory method for evaluating vehicle seat vibration — Part 1: Basic requirements*

3 Terms, definitions and symbols

3.1 Terms and definitions

For the purposes of this document, the terms and definitions given in ISO 2041 and ISO 5805 apply.

ISO and IEC maintain terminological databases for use in standardization at the following addresses:

— ISO Online browsing platform: available at https://www.iso.org/obp

— IEC Electropedia: available at http://www.electropedia.org/

3.2 Symbols (units)

$a_z(t)$	input acceleration in z-direction depending on time (1 m/s^2)
$a_z(\omega)$	Fourier transform of $a_z(t)$ (1 m/s)
$A_z(t)$	time dependent spinal acceleration response function (1 m/s^2)
$A_z(\omega)$	frequency dependent spinal acceleration response function (1 m/s)
$A_{z,i}$	i^{th} maximal value of $A_z(t)$ (1 m/s^2)
B	endplate area of a vertebra (1 mm^2)
C_{dyn}	response function of compressive force in Annex A (1 N)
$C_{dyn,i}$	i^{th} maximal value of C_{dyn} (1 N)
D_z	acceleration dose depending on $A_{z,i}$ for t_m (1 m/s^2)
D_{zd}	daily acceleration dose extrapolated for t_d (1 m/s^2)
$H(\omega)$	transfer function (1)
m_z	acceleration–compressive stress conversion factor depending on mass in Annex C [$1{\cdot}10^6$ Pa/(m/s^2) = 1 MPa/(m/s^2)]
N	number of exposure days per year (1)
Π	risk of vertebral failure, based on R (1)
S_d	daily compression dose for model in Clause 5 (1 MPa)
S_{stat}	static stress for model in Clause 5 (based on gravitation) (1 MPa)
$S_{u,i}$	vertebral ultimate strength for model in Clause 5 for year i (1 MPa)
S^A	compressive dose in Annex A (1 MPa)
S_d^A	daily compressive dose in Annex A (1 MPa)
S_q^A	compressive dose for variable exposures in Annex A (1 MPa)
S_{stat}^A	static stress for model in Annex A (based on mean C_{dyn}) (1 MPa)
$S_{u,i}^A$	vertebral ultimate strength for model in Annex A for year i (1 MPa)
t	time (1 s)
t_d	duration of daily exposure (1 s)
t_m	measurement duration (1 s)
ω	angular frequency (1 Hz)

R stress variable for the risk calculation for model in Clause 5 (1)

R^A risk factor based on S_d^A (1)

R_q^A risk factor based on S_q^A (1)

NOTE The quantities that describe the injury risk are defined in Annex C (model of Clause 5) and Annex E (model of Annex A). For Clause 5, the injury risk is described by $\Pi(R)$, which is a function of R. This stress variable R differs from the injury risk R^A for the model of Annex A, which is defined in Annex E.

4 Delineation of the two exposure regimes

The exposure conditions in this document differ from those for the basic evaluation of whole-body vibration as described in ISO 2631-1.

NOTE 1 ISO 2631-1:1997, Clause 6 contains criteria, when additional methods of evaluation need to be used, including ISO 2631-5.

There are two exposure regimes that have to be distinguished:

a) On the one hand, one finds severe conditions which are typical for military off-road vehicles or high speed marine craft, etc. These severe conditions can contain periods of free fall, they are dominated by accelerations in the z-axis, and the subjects can lose contact with the seat surface due to the exposure. These conditions are addressed in Clause 5 and in Annexes C and D. Here, the requirements for the measurement (bandwidth, signal conditioning) differ from those in ISO 2631-1, and the contributions of the x- and y-directions to the compressive forces in the spine are neglected since the exposure is dominant in the z-direction.

   NOTE 2 Issues arising from the limitation to a default posture and a purely vertical excitation are addressed in the Introduction and in Annex B.

b) On the other hand, less severe conditions are also covered by this document without free-fall events and where the subject remains seated throughout the measurement. These are more likely in an industrial context, e.g. driving with tractors, forestry machines and mobile earth-moving machinery over rough surfaces (off-road, potholes, frequent crossing of railroad tracks, etc.). These conditions are addressed in Annexes A and E. The requirements for the measurement are the same as for the unweighted acceleration time series described in ISO 2631-1.

   To determine the regime for a given exposure, two questions have to be used:

   i) Does the driver lose contact with the seat (or would the driver lose contact in absence of a restraint system)?

   ii) Does the exposure contain periods of free fall?

If either question is answered with yes, the method of Clause 5 and Annexes C and D has to be used.

In case of doubt, these criteria can be checked quantitatively by measuring a representative exposure with the method outlined in Clause 5 or in Annex A (the more likely one is chosen). The measured time series in z-direction at the person are then checked: after applying the band-limiting filters described in ISO 2631-1, the peak accelerations shall not exceed 9,81 m/s^2 for the use of Annexes A and E. If the peak accelerations thus obtained exceed 9,81 m/s^2, Clause 5 and Annexes C and D apply.

1) If one starts with the requirements for the severe conditions, one proceeds with the signal conditioning process up to the step before the band-limiting filter is applied. The check is performed with a copy of the signal, so that the correct band-limiting filter (see 5.1.3) can be applied after the successful check.

2) If one starts with the requirements for the less severe conditions, and if the check is successful, one can use the band-filtered signal from the check for the further evaluation.

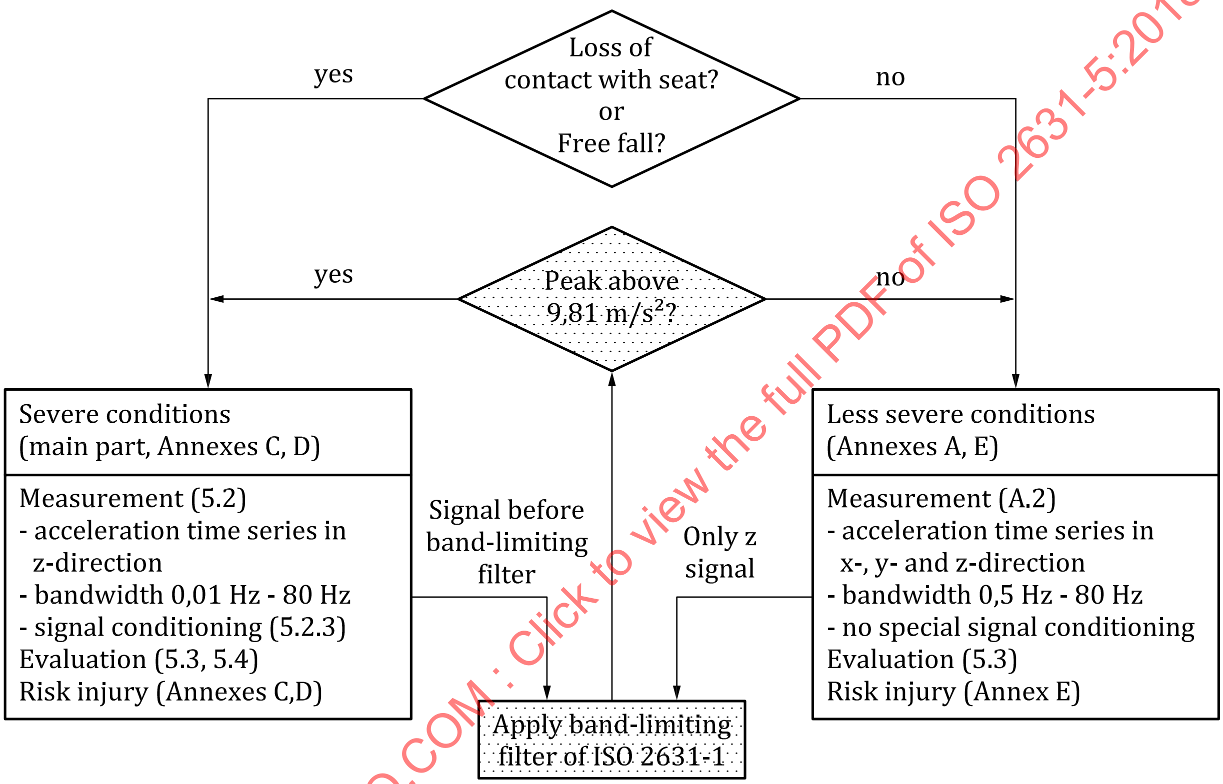
3) If the check does not confirm the first assumption of the exposure conditions, the measurement has to be repeated with the requirements of the other exposure condition.

This check is optional and, therefore, shaded in Figure 1.

NOTE 3 The band-limiting filters for the z-direction in ISO 2631-1 are:

— High pass: two-pole filter with Butterworth characteristic, corner frequency 0,4 Hz,

— Low pass: two-pole filter with Butterworth characteristic, corner frequency 100 Hz.



NOTE The shaded parts allow for an optional, quantitative confirmation of the first decision.

Figure 1 — Flowchart for the application of the models used in this document

5 Description of the model

5.1 Vibration measurement

5.1.1 General considerations

Vibration measurement, including the direction of measurement, location of transducers, duration of measurement, and reporting of vibration conditions, shall follow the requirements given in ISO 2631-1 except as described in 5.1.2 to 5.1.4.

5.1.2 Measurement location and specific hardware requirements

The vertical acceleration $a_z(t)$ should be measured at the interface between the seat and the ischial tuberosities.

During data collection, the subject should remain seated and should not rise from the seat. The location of measurements on the seat and the design of the accelerometer disk on the seat pad shall be as specified in ISO 10326-1.

Contact switches, video recordings or other methods should be used to detect loss of contact between the subject and the seat surface. It is necessary to detect and report the loss of contact, since accelerations measured during loss of contact shall not be counted as exposure. In addition, it shall be ensured that the impact experienced landing on the seat (i.e., both the motion of the person *and* the motion of the seat) after free fall is fully taken into account.

The accelerometers and associated measuring equipment shall be appropriate for measuring the highest amplitude accelerations anticipated during the measurement period.

The recorded, digitized accelerations should have a flat acceleration frequency response from 0,01 Hz to at least 80 Hz. A sampling rate of 256 samples per second or greater can be necessary depending on the anti-aliasing method used.

Details of the measurement equipment, including description of the calibration methods used, shall be provided.

5.1.3 Signal conditioning

The different steps of the signal conditioning process are summarised in Figure 2.

1. Sign check (cranial +)
2. Elimination of loss of contact
3. Resampling
4. Offset correction
5. Tapering
6. Band-limiting filter

Figure 2 — Steps of the signal conditioning process

In the first step, it is important to check that the sign of acceleration signals (positive, negative) is correct as the analysis method is concerned with compressive spinal loading. In the basicentric coordinate system for seated persons, the direction of the z-axis acceleration is positive to cranial (i.e. upward is positive).

After the sign of the acceleration signals has been checked, the second step eliminates those parts of the signal where there is no contact between the accelerometer disk on the seat pad and the subject. This leads to separate parts of the signal to which the following steps are applied separately.

In the third step, if the data have to be re-sampled for analysis after being acquired at a higher frequency, then it is necessary to check that appropriate anti-aliasing filtering is used.

NOTE 1 Resampling functions provided by common data processing software packages, such as MATLAB®[1)], can apply suitable anti-aliasing filters automatically but it is important to check that this is the case.

In the fourth step, the measured acceleration should have an offset correction so that the recorded acceleration, with the accelerometer at rest (or with a symmetric signal), is (0 ± 0,1) m/s^2. Note that subtraction of the mean may not be appropriate if the recorded acceleration is asymmetric.

In the fifth step, if analysing a time history where the accelerometer was in motion at the start or end of the recording, then tapering the signal, for instance with a cosine taper applied over several seconds, may be appropriate before applying the band-limiting filters.

1) MATLAB® is the trademark of a product supplied by MathWorks. This information is given for the convenience of users of this document and does not constitute an endorsement by ISO of the product named.

Finally, in the sixth step, the offset-corrected acceleration measurements shall be band-limited at 0,01 Hz and 80 Hz using second order Butterworth high pass filter with cut-off frequencies of 0,01 Hz and a fourth order Butterworth low pass filter with a cut-off frequency of 80 Hz. The band-limiting and weighting filters described in ISO 2631-1 should not be applied.

NOTE 2 The low frequency limit is reduced from that in ISO 2631-1 to prevent distortion of the acceleration signal if there is a period of free fall before a severe impact. Free-fall periods in excess of 0,5 s have been observed for fast naval craft. High pass filtering at 0,5 Hz causes the –1 g free-fall acceleration to be shifted back to zero by the time the impact occurs. This causes the peak acceleration of the impact to be incorrectly offset by up to +1 g. Calculations with a limited number of fast naval craft motions suggested errors of 10 % could be caused. Abrupt changes in terrain contour or steep slopes can cause a similar effect. If there is little movement at the measurement location at frequencies below 0,5 Hz, then errors caused by distortions of the time history due to the filter are likely to be small.

5.1.4 Measurement duration

The duration of the measurement shall be sufficient to ensure that measured results are representative of the exposure, i.e. that the measured multiple shocks are typical of the exposures that are being assessed.

Since shock events may be infrequent, consideration should be given to estimating the duration required to obtain a sufficient number of representative impacts. The duration of measurement should be appropriate to the assessment of the overall exposure.

NOTE 1 It is not practical to specify a sufficient number of impacts as this depends on how variable the impacts are. If the severity of impacts is variable with some at relatively low magnitudes and a few severe shocks, then a longer measurement is likely to be necessary to increase the probability of capturing these severe shocks.

NOTE 2 In repeatable tasks (e.g. mine haul trucks), recording at least a complete work cycle would be representative. In non-repeatable tasks (e.g. off-road travel, military transport), the sufficient duration depends also on the variability in the terrain.

Careful consideration shall always be given to controlling the shock and vibration exposure of any personnel involved in the trial. It may be appropriate to take shorter measurements initially to gain confidence that exposures will not be excessive. Trials where humans are exposed to repeated shock are likely to require careful risk assessment.

5.2 Determination of spinal response

A seat-to-lumbar spine transfer function has been developed based on experimental results and numerical modelling of seated occupants[4][5][15][72]-[75].

The frequency response of the transfer function between the seat and the spine is given in Formula (1) in terms of one complex zero and six complex poles.

$$H(\omega)=\frac{1+2\zeta_1\frac{j\omega}{\omega_1}+\left(\frac{j\omega}{\omega_1}\right)^2}{\left\{\begin{array}{c}\left[1+2\zeta_2\frac{j\omega}{\omega_2}+\left(\frac{j\omega}{\omega_2}\right)^2\right]\left[1+2\zeta_3\frac{j\omega}{\omega_3}+\left(\frac{j\omega}{\omega_3}\right)^2\right]\\ \left[1+2\zeta_4\frac{j\omega}{\omega_4}+\left(\frac{j\omega}{\omega_4}\right)^2\right]\left[1+2\zeta_5\frac{j\omega}{\omega_5}+\left(\frac{j\omega}{\omega_5}\right)^2\right]\\ \left[1+2\zeta_6\frac{j\omega}{\omega_6}+\left(\frac{j\omega}{\omega_6}\right)^2\right]\left[1+2\zeta_7\frac{j\omega}{\omega_7}+\left(\frac{j\omega}{\omega_7}\right)^2\right]\end{array}\right\}} \tag{1}$$

where

j $=\sqrt{-1}$;

ω_1 = 34 rad/s;

ζ_1 = 0,35;

ω_2 = 31 rad/s;

ζ_2 = 0,21;

ω_3 = 230 rad/s;

ζ_3 = 0,88;

ω_4 = 260 rad/s;

ζ_4 = 0,80;

ω_5 = 320 rad/s;

ζ_5 = 0,40;

ω_6 = 380 rad/s;

ζ_6 = 0,75;

ω_7 = 420 rad/s;

ζ_7 = 0,65.

Tolerances for any implementation of this transfer function by analogue or digital means are

a) ±2,5 % of the peak magnitude (i.e. ±0,04) about the target magnitude response and ±π/(10 rad) about the target phase response from zero to 40 Hz;

b) ±5 % of the peak magnitude (i.e. ±0,08) about the target magnitude response from 40 Hz to 80 Hz (no phase requirement);

c) ±5 % of the peak magnitude (i.e. +0,08) about the target magnitude response above 80 Hz (no lower bound and no phase requirement).

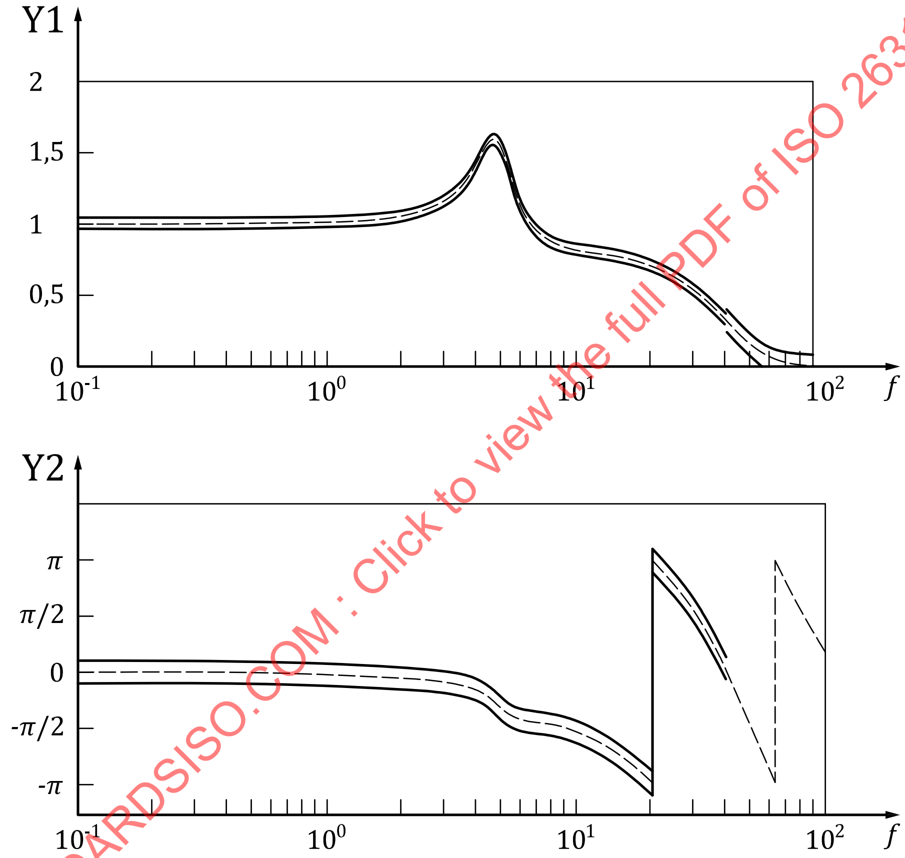
The frequency response of the seat to spine transfer function with tolerances is shown in Figure 3.

An example of a digital implementation of this frequency response function for a digital time history sampled at 256 samples per second is given in Annex D.

This transfer function of Formula (1) shall be applied to the Fourier transform of the measured conditioned vertical acceleration $a_z(\omega) = F[a_z(t)]$ following signal conditioning as described in 5.1.3.

$$A_z(\omega) = H(\omega) a_z(\omega)$$
$$A_z(t) = F^{-1}\left[A_z(\omega)\right] \tag{2}$$

This gives rise to $A_z(\omega)$ which is the response function of the spine in the frequency domain (spectrum). The response function of the spine in the time domain $A_z(t)$ is given by the inverse Fourier transform of $A_z(\omega)$.



Key

f frequency, Hz

Y1 magnitude

Y2 phase, rad

Figure 3 — Frequency response of the seat to spine transfer function with tolerance bands

NOTE This response was derived from a linearization of the neural network[15][72]-[75]. A least squares fit was applied to the neural network response to a Gaussian random DC to 80 Hz signal with a peak acceleration of 20 m/s^2. The resulting filter was adjusted to give a transmissibility at DC of 1,0 on the basis that the non-unity transmissibility at DC of the neural network was physically incorrect. Additional attenuation was applied above approximately 50 Hz to provide an upper limit for the frequency response. Confidence in the original neural network response frequencies over 50 Hz was considered to be low due to the considerable uncertainties associated with measuring the transfer functions.

5.3 Calculation of spinal response dose

The acceleration dose D_z, in metres per second squared, in the vertical direction is defined as

$$D_z = 1{,}07 \left(\sum_i A_{z,i}^6 \right)^{\frac{1}{6}} \tag{3}$$

where $A_{z,i}$ is the i th peak of the response acceleration $A_z(t)$ [Formula (2)].

NOTE 1 Formula (1) represents the frequency response of the spine in terms of a filter. In addition, there is an amplitude response function of the spine, which is represented here by the factor of 1,07 in Formula (3).

A peak is defined here as the maximum value of the response acceleration between two consecutive zero crossings. Only positive peaks shall be counted.

In calculating the dose, acceleration peaks of a considerably lower (by a factor of three or more) magnitude than the highest peak do not significantly contribute to the value associated with the 6th power term in Formula (3).

A daily dose, D_{zd}, in metres per second squared, can be calculated using Formula (4):

$$D_{zd} = D_z \left(\frac{t_d}{t_m} \right)^{\frac{1}{6}} \tag{4}$$

where

D_z is the acceleration dose;

t_d is the time period of the daily exposure;

t_m is the time period over which D_z has been measured.

To be able to compare different exposures, it is recommended to use t_d = 8 h in Formula (4).

Formula (4) may be used when the total daily exposure can be represented by a single measurement period. When the daily vibration exposure consists of two or more periods of different magnitudes, the acceleration dose, in metres per second squared, for the total daily exposure may be calculated as follows:

$$D_{zd} = \left(\sum_j D_{z,j}^6 \frac{t_{d,j}}{t_{m,j}} \right)^{\frac{1}{6}} \tag{5}$$

where

$D_{z,j}$ is the acceleration dose for condition j;

$t_{d,j}$ is the duration of the daily exposure to condition j;

$t_{m,j}$ is the duration over which $D_{z,j}$ has been measured.

NOTE 2 Guidance on the assessment of adverse health effects from the knowledge of the spinal shocks is given in Annexes B and C.

Annex A
(informative)

Alternative model for the determination of spinal response during exposures without loss of contact with seat surface

A.1 General

The focus of Clause 5 is on exposures in severe conditions: military off-road vehicles, high speed marine craft, etc. However, this annex offers an alternative model for the assessment of multiple shocks for exposures where the exposed subject does not lose contact with the seat surface due to the shock. These exposures are more likely to be seen in an industrial context, e.g. the exposures while driving tractors, trucks, forestry machines and mobile earth-moving machinery. A connection between the exposure assessment of this annex and adverse health effects has been established in one study[69][71].

This alternative model is validated for peak accelerations up to 9,81 m/s^2 for acceleration signals in z-direction measured at the seat surface. Therefore, this annex is restricted to measurements where the exposed person does not lose contact with the seat surface due to shocks. The bandwidth of the measurement and the signal conditioning according to Clause 5 accounts for free-fall events, which are not expected in the industrial context of this alternative model. Therefore, the measurement can follow the procedures outlined in ISO 2631-1. Detailed information on the measurement of acceleration time series is given in A.2.

The alternative model processes as input acceleration signals in three directions measured at the seat surface (minimum). In addition, measurements at the backrest and the feet (seat mounting point or cabin floor) can also be included in the evaluation. On the basis of these time series, the compressive forces between two vertebrae are calculated with the help of transfer functions of a biomechanical model, which are included in a calculation software, available at http://standards.iso.org/iso/2631/-5/ed-2.

The user is permitted to use this calculation software in its original format without any modifications for the purposes specified in the document. The usage of the calculation software is described in the user instruction provided at the same server address.

Therefore, the effect of accelerations in all three directions is taken into account, while in Clause 5 the assessment is based only on the z-direction. In addition, the transfer functions depend on the posture, body mass and body mass index (BMI) of the exposed individual. While the body mass is also a variable in Annex C (see Formula C.1), the assessment is restricted there to a default posture (upright sitting).

NOTE 1 The BMI is the ratio of the body mass (kg) and square of the body height (m).

Given the measured acceleration and exposure conditions, the software calculates by means of transfer functions the spinal response in terms of compressive forces (see A.3.2). The peak compressive forces thus obtained give rise to a compressive dose defined in A.3.3.

NOTE 2 Guidance on the assessment of adverse health effects on the basis of this compressive dose is given in Annexes B and E.

A.2 Measurement of acceleration time series

Vibration measurement, including the direction of measurement, location of transducers, duration of measurement, and reporting of vibration conditions, shall follow the requirements included in ISO 2631-1 and also ISO 10326-1 for the location of measurements on the seat and for the design of the mounting disc.

The software (see A.3) evaluates only the signal portion between 0,5 Hz and 80 Hz. Therefore, the sampling rate has to be larger than 160 Hz. Vibration transducers should register adequately accelerations from 0,5 Hz to 80 Hz. Data should be acquired effectively synchronously across all directions and measurement locations.

This alternative model is validated for peak accelerations up to 9,81 m/s^2 for acceleration signals in z-direction measured at the seat surface, which are filtered by the band-limiting filters described in ISO 2631-1 (high pass: two-pole filter with Butterworth characteristic, corner frequency 0,4 Hz; low pass: two-pole filter with Butterworth characteristic, corner frequency 100 Hz). For assessment, the band-pass filtered signal from the seat surface generated to test for the maximal peak accelerations can be used. No further signal conditioning is needed. For the other measurement locations (backrest, seat mounting point or cabin floor), frequency-weighting filters according to ISO 2631-1 should not be applied.

For measurement of vibrations including multiple shocks, it is important that the sign of acceleration signals (positive, negative) is correctly recorded. In the basicentric coordinate system for seated persons, the direction of the y-axis is positive to the subject's left. The direction of the x-axis is positive to ventral, and the direction of the z-axis is positive to cranial.

During data collection, the subject shall remain seated and belted, if possible, and shall not lose contact with the measurement disc.

In addition to the measurement (at least) at the seat surface, measurements at the backrest are recommended. To account for the exposure at the feet and hands, it is recommended to measure accelerations at the seat mounting point or cabin floor.

The duration of the measurement shall be sufficient to ensure that the multiple shocks are typical of the exposures that are being assessed, without risk of injury to the exposed individuals during the measurement. Careful consideration should be given to accurate characterization of the statistical distribution of the impacts since the higher-amplitude tails of the distribution might not occur during the sampling duration but can substantially affect potential injury assessments.

The time series shall be saved in separate files for each interface of the model: seat surface, backrest, hands, and feet. This requires saving two copies of the measured signal from the seat mounting point or cabin floor: one for the hands and one for the feet. All files shall not contain a header line and the data shall be in four columns: 1st for measurement time in seconds, 2nd for acceleration in x-direction in m/s^2, 3rd for acceleration in y-direction in m/s^2, 4th for acceleration in z-direction in m/s^2.

A.3 Software

A.3.1 Software input and output (orientating analysis)

The data files and paths of the measured acceleration time series (A.2) have to be specified in the input file of the software. If there are no measurements available from the backrest, this can be substituted with the measurements at the seat surface. The exposure at the feet and hands is represented by one measurement at the seat mounting point or cabin floor. If only data from the seat surface are available, this is used for the seat and backrest and "Zero" is indicated for the other measurement points. This assumes a constant value of 0 m/s^2 for the hands and the feet.

In addition, the model needs as input the posture, body mass and height (BMI) of the exposed individual. Also the life-time exposure history has to be provided. In order to compare different exposures or to make an assessment when no additional information on the individual exposure is available, the orientating analysis of this annex uses as an input the same set of conditions (default values) and normalizes in particular the measured exposure to a typical/realistic daily exposure duration of 4 h. The default values are chosen in such a way that they maximise the spinal load:

a) a high BMI and the highest body mass percentile possible (BMI > 26,1 kg/m^2 and 95th percentile, i.e. a body mass larger than 109 kg),

b) the exposure lasts from the age of 20 to 65 years for 240 days per year normalized to 4 h per day,

c) unfavourable driver's posture [group 3 (excavator), see Table A.1 and Figure A.1].

The postures (see Table A.1 and Figure A.1) are represented in the software by mean values of body angles for drivers using five different vehicle types: (1) fork lift truck, (2) wheel loader, (3) excavator, (4) forwarder, (5) harvester[27].

NOTE 1 In the individual analysis in Annex E, different BMI, daily and life-time exposure durations and postures can be used. All five posture groups are used in Annex E, including posture group 3.

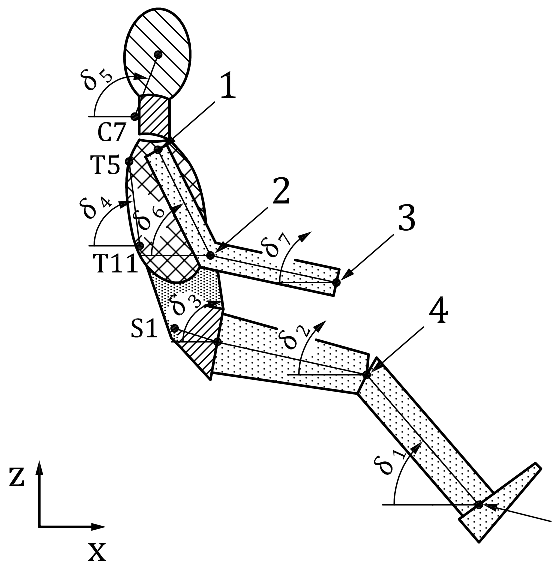
On the basis of the measured accelerations and the default values, the analysis software calculates the intervertebral compressive forces (see A.3.2) and the maximum daily compressive dose (max. S_d^A) for the six disc levels of the lumbar spine [Formula (A.3) with j = 1]. Although the measurement duration $t_{m,1}$ in Formula (A.3) is not restricted, it is assumed that the daily exposure duration is normalized in the orientating analysis to 4 h [$t_{d,1}$ = 4 h in Formula (A.3)]. Individual exposure durations for the measured exposures can be accounted for in Annex E (individual analysis). The results as well as the input information given for the analysis are written in a spreadsheet (output file).

NOTE 2 In addition, the software calculates the risk factors R^A for the six disc levels based on S_d^A according to Annex E, writes this additional information in the output file and shows it on the screen.

Table A.1 — Angle values for posture groups

Angles in degrees

Model/posture group	δ_1	δ_2	δ_3	δ_4	δ_5	δ_6	δ_7
Group 1	72,0	7,3	18,0	92,7	129,3	48,7	-6,0
Group 2	68,0	11,8	18,0	89,5	119,5	43,7	25,5
Group 3	64,0	2,0	18,0	100,0	125,5	75,4	6,1
Group 4	75,5	8,0	18,0	89,7	121,7	75,8	9,3
Group 5	84,7	7,9	18,0	93,2	122,0	76,9	13,0
NOTE See Figure A.1.							



Key

1 shoulder
2 elbow
3 hand
4 knee
5 foot

NOTE Values for posture groups are given in Table A.1.

Figure A.1 — Sketch of postural angles

A.3.2 Calculation of intervertebral forces

The internal lumbar forces for the excitation time-histories measured according to A.2 are calculated on the basis of transfer functions which have been derived from the results of a group of Finite Elements (FE) models of the seated human. These FE models present posture variations of an anatomy-based FE model for the upright seating posture considering the impact of different acceleration magnitudes[29][31][49][50]. For this posture, a large number of measurements for the apparent mass and the transfer functions to different body parts are available[26] on the basis of which the model was validated intensively. Particular attention was paid to the modelling and validation of the lumbar spine and the associated musculature. For example, the modelling and validation of the lumbar spine was carried out stepwise from the vertebral body and vertebral disk separately via individual motion segments to the entire lumbar spine. The FE model also contains the effect of muscular forces by including passive models for the muscle forces that maintain the posture. These forces are added to the dynamic compressive forces in the vertebral discs.

This aspect, along with the strict orientation on human anatomy when modelling the other body parts and joints, allows for the adaptation of the model to typical working postures and typical drivers' anthropometries. For application within the framework of Annexes A and E, FE models were generated for five common and typical working postures with ten representative drivers' anthropometries each, which depend on the BMI[27]. In this annex, only one posture and one anthropometry are used (see A.3.1). By means of these models, the transfer functions for accelerations of different acceleration magnitude classes from the four human-machine interfaces (buttock, back, feet and hands) to the forces (compression and shear) in the lumbar spine were calculated[57].

All further calculations are carried out by means of these transfer functions. The software automatically assigns the acceleration magnitude class from the measured accelerations (A.2), and it also assigns the drivers' anthropometry from the provided height and weight of the driver.

Based on the transfer functions $h_{ed,el,fd}^{dl}$ and the Fourier transforms of the measured excitation time series $a_{ed,el}$ (Formula A.1), the software calculates the spinal forces for each excitation location *el* (human-machine interface) and excitation direction *ed* in the frequency domain. Summation over the excitation locations and directions gives rise to the compression and shear forces f_{fd}^{dl} between the vertebrae due to the combined excitation:

$$f_{fd}^{dl} = \sum_{el=1}^{4} \sum_{ed=1}^{3} h_{ed,el,fd}^{dl} \cdot a_{ed,el} \tag{A.1}$$

where

f_{fd}^{dl} is the vector of forces at disk level $dl \in \{\text{T12-L1}, \cdots, \text{L5-S1}\}$. Since only the compression-decompression force (cd) is used in this document, the force direction fd = cd.

$h_{ed,el,fd}^{dl}$ is the transfer function from accelerations at excitation location $el \in \{buttock = 1, back = 2, hands = 3, feet = 4\}$ in direction $ed \in \{x = 1, y = 2, z = 3\}$ to forces at disc level *dl* for force directions *fd*;

$a_{ed,el}$ is the Fourier transform of the acceleration at excitation location *el* in direction *ed.*

All quantities in Formula A.1 are complex and functions of the frequency. Finally, the complex, frequency dependent compression-decompression forces f_{fd}^{dl} are transferred into the time domain by an inverse Fourier transformation. This results in time dependent compression-decompression forces C_{dyn} (N) which are used in a subsequent step to calculate a daily compressive dose value S_{d}^{A} (A.3.3).

NOTE 1 The life-time exposure in combination with S_{d}^{A} is needed for the risk factor R^{A} in Annex E.

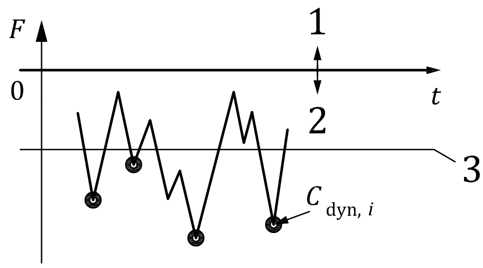
NOTE 2 For Gaussian random signals, it is possible to calculate the compressive dose S_{d}^{A} within the frequency domain[76]-[78].

A.3.3 Calculation of a compressive dose

Since the Palmgren-Miner fatigue theory[12][13][17][21][26][33]-[37][40][41][46] only refers to those fatigue fractures caused by the compressive forces, only peaks which cause compression at the end plates shall be counted (compression of the spine is of primary interest for exposure severity).

NOTE So far, there is no reliable information available about the damage mechanisms caused by shear forces.

The compressive dose S^{A} (MPa) per exposure is defined as the sum of peak compressive forces $C_{\text{dyn},i}$ (N) acting on the area of a vertebral endplate B (mm^2). These three quantities are determined separately for every disc level. The peak compressive forces are negative values and defined here as a maximum value of the additional compressive force between two consecutive mean value crossings (see Figure A.2).



Key

1 tensile side

2 compressive side

3 mean value = static compressive force

F force

Figure A.2 — Sketch for the calculation of $C_{dyn,i}$

$$S^A = \left[\sum_i \left(\frac{C_{dyn,i}}{B} \right)^6 \right]^{\frac{1}{6}} \quad \text{(A.2)}$$

The vertebral endplate areas are already incorporated in the software on the basis of literature data[62]. Average values for B are for disc-level T12/L1 1 460 mm^2, L1/L2 1 520 mm^2, L2/L3 1 580 mm^2, L3/L4 1 590 mm^2, L4/L5 1 600 mm^2, L5/S1 1 550 mm^2.

Experimental data show that the value of the Palmgren-Miner exponent varies with biological tissue and test methodology from 5 to 14 for cortical and trabecular bone to 20 for cartilage. For the purpose of estimating adverse health effects, a conservative exponent of 6 has been chosen[15][16] in Formulae (A.2) and (A.3). Due to the 6th power term, compressive force peaks of a considerably lower (by a factor of three or more) magnitude than the highest peak do not significantly contribute to the compressive dose.

In order to estimate health effects, it is useful to determine the equivalent daily compressive dose S_d^A (MPa) of the lumbar spine which considers j exposures during a day:

$$S_d^A = \left(\sum_j S_j^{A\,6} \frac{t_{d,j}}{t_{m,j}} \right)^{\frac{1}{6}} \quad \text{(A.3)}$$

where

S_j^A is the dynamic compressive stress of the lumbar spine due to vibration exposure to condition j;

$t_{d,j}$ is the time period of the daily vibration exposure to condition j;

$t_{m,j}$ is the time period over which S_j^A has been measured.

The analysis of this annex compares a single exposure (j = 1) and assumes $t_{d,1}$ = 4 h.

Annex B
(informative)

General relationship between acceleration dose and health effects

This annex applies to people in normal health who are regularly exposed to vibration containing multiple shocks. Individuals with previous disorders affecting the spine, including those suffering from latent osteoporosis or other spinal disorders can be more susceptible to injury.

It is assumed that multiple shocks cause transient pressure changes at the lumbar vertebral endplates that over time can result in adverse health effects, arising from material fatigue processes. Essential exposure-related factors are the number, magnitudes and rise times of peak compression in the spine. The peak compression in the spine is affected by anthropometric data (body mass, body height, size of endplates) and posture[27][35][58]. Adverse health effects of long-term whole-body multiple-shock exposure include an increased risk to the lower lumbar spine and the connected nervous system of the segments affected. Excessive mechanical stress and/or disturbances of the nutrition of and diffusion to the disc tissue can contribute to the degenerative processes in the lumbar segments[32]-[35][57][58]. Multiple-shock and vibration exposure can also worsen certain endogenous pathological disturbances of the spine.

The relationship between the predicted pressure changes and the predicted total tolerance of the exposed person can be used to assess the potential of an adverse health effect. The predicted response is of the bony vertebral endplate (hard tissue). A bending forward or twisting posture is likely to increase the adverse health effect.

The intervertebral disc and paraspinal ligaments and muscles (soft tissue) can be at risk of injury in multiple mechanical shock environments for the following reasons, which have been documented[3][14][18]-[20][23][43][45][51][53][59]-[65]:

a) The seated posture can be mechanically stressful on the disc.

b) Different postures can change the way the body responds to multiple loads, inconsistent with the model constraints.

c) The intervertebral disc can change internal pressure, soften, tear and/or buckle with exposure to multiple loads.

d) The condition of the intervertebral motion segment depends on the proper functioning of the neuromuscular control system for active and passive stabilization, and therefore to prevent buckling.

e) Impact can be uncomfortable, can be considered an unexpected, sudden load and can lead to an overcompensating response in the trunk muscles.

f) Impact, especially following multiple load exposure, can trigger a buckling event in the intervertebral motion segment due to an inability of the neuromuscular control system to respond fast enough in a coordinated fashion.

Annex C
(informative)

Assessment of health effects for exposures that are described in Clause 5

By use of a biomechanical model, based on experimental data, it has been shown[15][71]-[75] that there is a linear relationship between the part of compressive stress that is due to the input shocks and the peak acceleration response in the spine. After calculating the acceleration dose D_{zd} for the average daily exposure time [see Formulae (4),(5)], the daily compression dose S_d is obtained by multiplying D_{zd} by m_z:

$$S_d = m_z D_{zd} \tag{C.1}$$

The value of m_z provides a conversion between acceleration and compressive stress on the vertebral body in the seated posture. This value represents the mass borne above the pelvis divided by the spine endplate area. For young populations, the percentage of body mass is 49 % for men[67][68]. For an 82 kg man and an endplate area of 14 cm^2, the value for m_z is:

$$m_z = (82\ \text{kg}\cdot 0{,}49)/(14\ \text{cm}^2) = 0{,}029\ \text{MPa}/(\text{m/s}^2) \tag{C.2}$$

NOTE 1 The equivalent estimate of Formula (C.2) for young women would be: 45 % of the body mass of a 64 kg woman and an endplate area of 11,6 cm^2 leads to m_z = 0,025 MPa/(m/s^2).

Whenever possible, m_z should represent the specific individual or population for which the exposure risk is being calculated. For the most conservative risk estimation, the highest body masses which may be exposed to the shock amplitudes should be used. When additional gear is worn on the torso or head (e.g. body armor, heavy equipment, helmets), that mass should be added to the body mass borne above the pelvis.

NOTE 2 In ISO 3411:2007, the 95th percentile of an earth-moving machinery operator has a mass of 114,1 kg. In ISO/TR 7250-2:2010, the 95th percentile male person in the US has a mass of 115 kg, and in Japan 84,0 kg.

In this annex, the injury risk is described as a function of a stress variable R [see $\Pi(R)$ in Formula (C.5)]. The stress variable can be defined for use in the assessment of the adverse health effects related to the human response acceleration dose. R should be calculated sequentially taking into account increased age (and reduced strength) as the exposure time increases. It is defined as follows:

$$R = \left[\sum_{i=0}^{n-1} \left(\frac{S_{d,i} N_i^{\frac{1}{6}}}{S_{u,i} - S_{stat,i}} \right)^6 \right]^{\frac{1}{6}} \tag{C.3}$$

where

N	is the number of exposure days per year;
i	is the year counter;
n	is the number of years of exposure ([n]=1);

$S_{d,i}$ is the daily compression dose related to year i;

$S_{stat,i}$ is a constant representing the static stress due to gravitational force which depends, for example, on the body mass for year i;

$S_{u,i}$ is the ultimate strength of the lumbar spine for a person of age $(b + i)$ years with b being the age at which the exposure started ($[b]=1$).

The value of S_{stat} should be equivalent to m_z multiplied by 9,81 m/s^2, i.e. for the values of Formula (C.2), S_{stat} = 0,029 MPa/(m/s^2) · 9,81 m/s^2 = 0,281 MPa.

The value $S_{u,i}$ varies with the strength of the vertebrae, which normally is reduced with age. From in-vitro studies[64], the following relationship between $S_{u,i}$ (in MPa) and $b + i$ (in years) has been derived:

$$S_{u,i} = 6{,}75\ \text{MPa} - S_{age}\,(b + i) \quad \text{(C.4)}$$

For n years of exposure, $0 \le i \le n{-}1$ in Formula (C.4). For male individuals, S_{age} is 0,052 MPa.

NOTE 3 For female individuals, S_{age} is 0,039 MPa.

NOTE 4 Formulae (C.4) and (E.2) have been derived in different ways and are, therefore, not identical.

To estimate the R value associated with the risk of lumbar spinal injury from axial loading, a survival analysis was performed[66] using the results of 107 human cadaveric lumbar spine fatigue tests from five different studies[12][22][24][25][34][42]. The specimens were both male (n = 78) and female (n = 29) and varied in age from 19 years to 93 years at death. In each of these tests, the specimen was a lumbar spinal segment composed of two vertebral bodies and the intervertebral disc with posterior elements and ligaments intact. The segments were fixed at each end and exposed to pure axial, cyclic compression ranging from 404 N to 7 100 N which corresponds to approximately 1,0 g to 18,1 g lumbar spinal acceleration based on an effective torso/head/upper extremity segmental mass of 40 kg based on a 82 kg male[52]. Since both the load magnitude and number of cycles varied between tested specimens, the R value was used to quantify the cumulative load. Determination of injury was made by the original authors. In the cases selected for this analysis, all injuries were classified as damage to the endplate or general bony structure.

Using a Weibull survival model[66], the probability of lumbar spine damage, Π, from axial load is calculated using the R value as follows:

$$\Pi = 1 - \exp\left[-\left(\frac{R}{\alpha}\right)^{\beta}\right] \quad \text{(C.5)}$$

where α and β are gender-dependent coefficients ($[\alpha] = [\beta] = 1$); coefficients for males and females are shown in Table C.1. Values of Π vary between 0 (0 % risk of injury) and 1 (100 % risk of injury). There is a significant human variability, and this model does not account for the effects of remodelling and healing in spinal injury that can have a substantial influence over the course of years. The R values for 10 %, 50 % and 90 % risk of injury are shown in Table C.2.

Table C.1 — Coefficients for Formula (C.5) (lower 95 %, upper 95 % confidence intervals)

	α	β
Male	1,613 (1,460; 1,809)	2,799 (2,168; 3,511)
Female[a]	0,959 (0,854; 1,093)	3,709 (2,509; 5,207)

[a] Values for female subjects are based on a smaller data set.

Table C.2 — *R* values for risk of injury

	R value (lower 95 %, upper 95 % confidence intervals) Risk of injury		
	10 %	50 %	90 %
Male	0,72 (0,58; 0,89)	1,42 (1,27; 1,57)	2,17 (1,91; 2,48)
Female[a]	0,52 (0,41; 0,67)	0,87 (0,77; 0,98)	1,20 (1,04; 1,38)

[a] Values for female subjects are based on a smaller data set.

Calculation of adverse health effects is performed as shown in Figure C.1.

Acceleration dose D_z [see Formula (3)]

↓

Normalizing D_z to the average daily exposure time the acceleration dose yields D_{zd} [see Formulae (4, 5)]

↓

Calculation of the daily equivalent compression dose S_d using Formulae (C.1) and (C.2)

↓

Calculation of stress variable R using Formula (C.3) and determination of risk of end-plate injury using Formula (C.5)

↓

Risk of injury < 10 %: low likelihood of an adverse health effect for cumulative exposure
Risk of injury > 50 %: high likelihood of an adverse health effect for cumulative exposure

Figure C.1 — Flowchart for assessment of adverse health effects

Assuming a high-impact working tenure of 120 days/year from age b = 20 to age $b + n$ = 20 + 20 = 40, during which individuals experience five 40 m/s^2 accelerations each day, yields a compressive stress S_d of 1,623 MPa for an 82 kg male, which leads to R = 1,22.

$$D_{zd} = 1{,}07\left[\sum_{i=1}^{5}\left(40\frac{\mathrm{m}}{s^2}\right)^6\right]^{\frac{1}{6}} \approx 55{,}97\frac{\mathrm{m}}{s^2}$$

$$S_d = m_z D_{zd} = 0{,}029\cdot 55{,}97\ \mathrm{MPa} \approx 1{,}623\ \mathrm{MPa}$$

$$R = \left\{\sum_{i=0}^{20-1}\left[\frac{1{,}62\ \mathrm{MPa}\,(120)^{\frac{1}{6}}}{6{,}75\ \mathrm{MPa} - 0{,}052\ \mathrm{MPa}\,(20+\mathrm{i})}\right]^6\right\}^{\frac{1}{6}} \approx 1{,}22$$

Using Formula (C.5) and Table C.2, this indicates a moderate adverse health effect (10 % < risk of injury < 50 %) for men when exposed to these conditions during the duration of the entire exposure period.

$$\Pi\,(R) = 1 - \exp\left[-\left(\frac{1{,}22}{1{,}613}\right)^{2{,}799}\right] \approx 0{,}37$$

NOTE 5 The same exposure leads to S_d = 1,40 MPa for a 64 kg female and an R = 0,97, which is associated with a high risk of injury (50 % < risk of injury < 90 %; see Table C.2).

Annex D
(informative)

Example of digital implementation of transfer function for exposures that are described in Clause 5

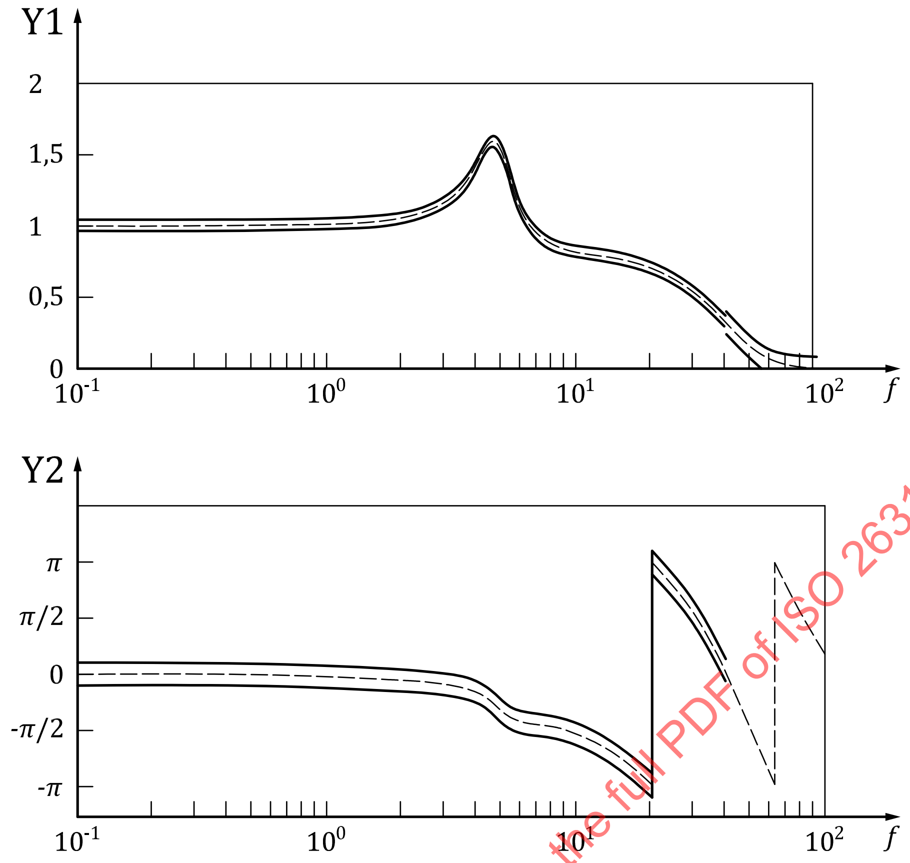
A digital time domain filter for the frequency response of the seat to spine transfer function [see Formula (1)] for an acceleration time history sampled at 256 samples per second can be obtained using Formula (D.1) as used by the “Direct Form II Transposed” filter function implemented in MATLAB® R2007a:

$$a(1)\cdot y(n)=b(1)\cdot x(n)+b(2)\cdot x(n-1)+\ldots+b(nb+1)\cdot x(n-nb)-a(2)\cdot y(n-1)-\ldots-a(na+1)\cdot y(n-na) \quad \text{(D.1)}$$

where $x(n)$ is the measured signal at step n and $y(n)$ is the filtered signal at step n. The coefficients $a(l)$ and $b(l)$ are given in Table D.1, and $na = nb = 11$. The response of this function is shown in Figure D.1.

Table D.1 — Digital filter coefficients for a sampling rate of 256 samples per second

l	*b*	*a*
1	−0,000 005 710	1,000 000 000
2	0,000 020 010	−3,323 217 600
3	0,001 373 900	4,256 126 150
4	0,014 541 920	−1,980 417 270
5	0,025 152 310	−1,488 735 470
6	−0,014 242 050	3,329 511 290
7	−0,044 262 840	−2,949 072 140
8	−0,008 888 510	1,653 403 410
9	0,017 715 720	−0,635 677 800
10	0,010 216 420	0,167 519 420
11	0,002 030 740	−0,028 076 980
12	0,000 055 980	0,002 348 730



Key

f frequency, Hz

Y1 magnitude

Y2 phase, rad

Figure D.1 — Digital filter magnitude and phase response with tolerances

Annex E
(informative)

Assessment of health effects for exposures without loss of contact with seat surface

E.1 General

The software of Annex A calculates a daily compressive dose S_d^A for every disc level on the basis of measured acceleration time series and default values for the individual exposure conditions (orientating analysis A.3.1). This annex is concerned with the so-called individual analysis which uses the same software described in Annex A. It differs from Annex A only in the choice of the individual exposure conditions (posture, BMI, life-time exposure duration, E.2), and in the calculation of the risk factor R^A for every disc level [Formulae (E.1),(E.4)], which describes, on the basis of S_d^A and the life-time exposure duration, the risk of spinal injury.

E.2 Software input and output (individual analysis)

The exposure accelerations have to be measured and provided in the same way as described in Annex A. By choosing the individual analysis in the software of Annex A, the user can combine different measurements of different machine(s), different posture(s) and BMIs, as well as different daily and life-time exposure durations. These data should be provided in the input file (spreadsheet):

a) year of birth of the exposed person;

b) first year of exposure;

c) last year of exposure;

d) duration of the several exposures/exposure pattern per day;

e) number of days of all exposures/exposure pattern per year;

f) body height (m) and body mass (kg) of the exposed person for each year with exposure.

If exposure patterns are changing over the years for the stress analysis of a working life and/or if there is a known physical change of the driver (e.g. increase of the body mass with increasing age), this information can be entered in the spreadsheet. Breaks in the life-time exposure history can be considered. If not all of the information above is available, the values of the orientating analysis can be used as default values, i.e. group 3 as default posture, BMI > 26,1 kg/m^2 and a body mass larger than 109 kg, exposure duration for 45 years beginning at age 20 for 240 days per year and normalized to 4 h per day.

The result of this individual analysis is the risk factor R^A [Formulae (E.1),(E.4)] for the vertebral levels T12/L1 to L5/S1 for an individual working life with varying exposure patterns and/or physical changes of the driver. Although R^A is calculated on the basis of S_d^A, the compressive daily dose values are not included in the output. The values for R^A are written in a spreadsheet (output file) and shown on the screen. In addition, the input information used for the analysis is also written in the output file.

The assessment should be carried out on the basis of the vertebral level with the highest R^A values.

E.3 Risk factor

In general, a risk factor R^A for every disc level can be defined for use in the assessment of an adverse health effect related to the daily compressive dose S_d^A described in A.3.3. The risk factor R^A considers the year at which the exposure started and the duration of exposure in relation to the age of the exposed person. For a constant exposure pattern per day in all years from the beginning to the end of exposure, the risk factor R^A is as follows:

$$R^A = \left[\sum_{i=1}^{n} \left(\frac{S_d^A N_i^{\frac{1}{6}}}{S_{u,i}^A - S_{stat,i}^A} \right)^6 \right]^{\frac{1}{6}} \tag{E.1}$$

where

S_d^A is the constant daily compressive dose;

i is the year counter;

N_i is the number of exposure days per year i;

n is the number of exposure years;

$S_{u,i}^A$. is the ultimate strength of a lumbar vertebra for a person of age (b+i) years with b being the age at which the exposure started;

$S_{stat,i}^A$ is the mean value of the compressive-decompressive force in Formula (A.1) divided by the area of a vertebra endplate B (mm^2)[62] for year i.

$S_{u,i}^A$ varies with the bone density of the vertebrae, which normally is reduced with age. From in-vitro studies, the following relationship between $S_{u,i}^A$ (MPa) and (b+i) (in years) has been derived, and the protection of 50 % of a driver population has been considered here[56][57]:

$$S_{u,i}^A = 6{,}765\ 024\ \text{MPa} - 0{,}067\ 184\ \text{MPa} \cdot (b+i) \tag{E.2}$$

NOTE 1 Formulae C.4 and E.2 have been derived in different ways and are, therefore, not identical.

The average daily compressive dose S_q^A for h variable exposure patterns per year is calculated in analogy to the compressive dose per day for every disc level:

$$S_q^A = \left(\sum_h S_{d,h}^{A\ 6} \frac{N_h}{N} \right)^{\frac{1}{6}} \tag{E.3}$$

where

$S_{d,h}^A$ is the constant daily compressive dose for exposure pattern h;

N_h is the number of days per year with exposure pattern h;

N is the number of days with exposure in the year.

$$R_{\mathrm{q}}^{\mathrm{A}} = \left[\sum_{i=1}^{n} \left(\frac{S_{\mathrm{q},i}^{\mathrm{A}} N_i^{1/6}}{S_{\mathrm{u},i}^{\mathrm{A}} - S_{\mathrm{stat},i}^{\mathrm{A}}} \right)^6 \right]^{1/6} \tag{E.4}$$

There is a significant human variability; $R^{\mathrm{A}} < 0{,}8$ indicates a low probability of an adverse health effect and $R^{\mathrm{A}} > 1{,}2$ indicates a high probability of an adverse health effect.

NOTE 2 The risk factor of Formula (E.1) cannot be interpreted as a probability of failure, i.e. $R^{\mathrm{A}} = 1$ does not correspond to a sure failure. It only indicates that the dynamic load due to mechanical shock has reached the same order of magnitude as the (static) ultimate strength that the vertebra is capable to resist.

For the exposure durations given in the orientating analysis (A.3.1), the risk factor for the individual disc levels can be calculated with the following parameters:

$$S_{\mathrm{d}}^{\mathrm{A}} = \left(S^{\mathrm{A}6} \frac{14\,400\ \mathrm{s}}{t_{\mathrm{m}}} \right)^{\frac{1}{6}} \tag{E.5}$$

$$R^{\mathrm{A}} = \left[\sum_{i=1}^{45} \left(\frac{S_{\mathrm{d}}^{\mathrm{A}} \cdot (240)^{\frac{1}{6}}}{S_{\mathrm{u},i}^{\mathrm{A}} - S_{\mathrm{stat},i}^{\mathrm{A}}} \right)^6 \right]^{\frac{1}{6}} \tag{E.6}$$

The calculation is based on 45 working years with 240 days of equal exposure (constant $S_{\mathrm{d}}^{\mathrm{A}}$) over 4 h (14 400 s).

A sequential calculation according to Formulae (E.1) or (E.4) with a constant static compressive stress $S_{\mathrm{stat}}^{\mathrm{A}}$ = 0,25 MPa for a person who starts being exposed at the age of 20 (b = 20) will reach R^{A} = 0,8 at the age of 65 (n = 45), if the daily compressive dose $S_{\mathrm{d}}^{\mathrm{A}}$ is equal to 0,5 MPa. The same person will reach R^{A} = 1,2 at the age of 65 if the daily compressive dose $S_{\mathrm{d}}^{\mathrm{A}}$ is equal to 0,75 MPa. This calculation is based on N = 240 days per year of equal exposure. For application to another number of days of exposure per year, the appropriate $S_{\mathrm{d}}^{\mathrm{A}}$ limits are achieved by multiplying the values 0,5 MPa and 0,75 MPa by $\left(\frac{240}{N} \right)^{\frac{1}{6}}$.

NOTE 3 This example illustrates how the values for R^{A} are calculated. The value of $S_{\mathrm{stat}}^{\mathrm{A}}$ = 0,25 MPa, however, is related to approximately 400 N for a disc area of 1 600 mm^2. This is a comparatively low value in[79], where static forces tend to be between 600 N and 1 000 N.

NOTE 4 When more experience of use of this document has been gained, comparison between these $S_{\mathrm{d}}^{\mathrm{A}}$ and R^{A} values and existing experience of adverse effects of long-term exposure might justify a re-evaluation of the values.

E.4 Transfer function for one set of conditions

This subclause contains an alternative access to the quantities presented in this annex without using the software. However, this is restricted to a fixed set of conditions as described below. To this end, a transfer function for the compressive-decompressive force is defined in Table E.1 and Figure E.1, which can be applied in analogy with Formula (A.1) to a given measured acceleration in z-direction. This results in the spinal compressive-decompressive force on the L4/L5 spine level. In addition, this force function has to be transferred into the time domain and the compressive dose $S_{\mathrm{d}}^{\mathrm{A}}$ (A.3) has to be calculated. The risk factor R^A according to Formulae (E.5) and (E.6) is calculated on the basis of 45 years of exposure starting at the age of 20 with a daily exposure duration of 4 h (orientating analysis A.3.1).

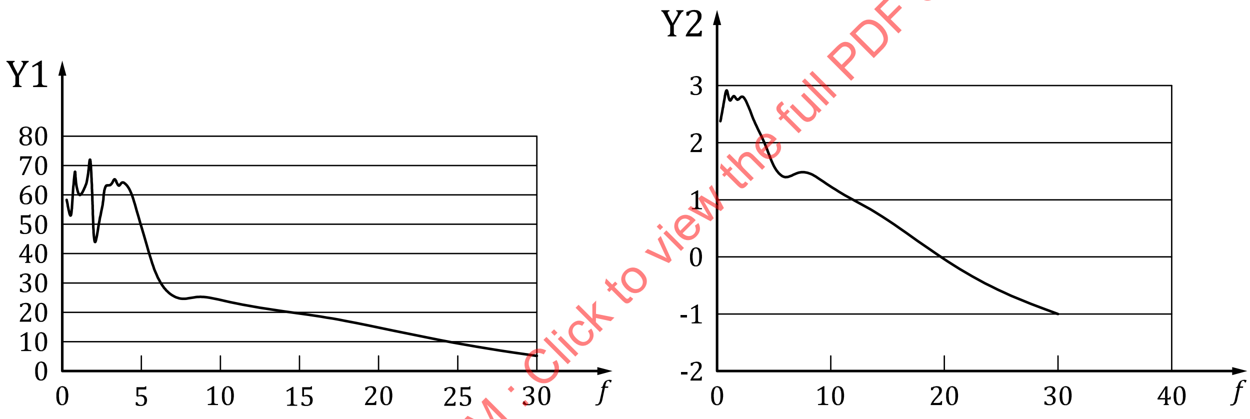
NOTE The transfer function in Table E.1 and Figure E.1 has only been constructed to allow for an alternative calculation of the *R*-value. It can only be used in the way presented in this subclause with respect to one signal in z-direction. It has been constructed to compensate for the effects of the horizontal directions. Therefore, it is not identical to any of the transfer functions of Formula (A.1).

The risk factor R^A obtained in this way should only deviate by 1 % to 2 % from the R^A -value calculated with the software for the following conditions:

1) identical acceleration a_z with an unweighted RMS value < 0,65 m/s^2 at buttock, back, hands and feet without excitation in the horizontal directions;

2) compressive-decompressive force on the L4/L5 spine level of a person with a body mass index larger than 26,1 kg/m^2 and a body mass equal to or above 109 kg in a sitting posture according to model group 3 (see Table A.1 and Figure A.1, typical e. g. for excavators).

Some deviations are to be expected, because even solely vertical excitation leads to little horizontal spine movements which contribute to the compressive-decompressive force. The transfer functions for the horizontal spinal forces are not considered in Figure E.1 and Table E.1. Depending on the input files, this can cause a lowering of the results in comparison with the analysis software when applying the code.



Key

f frequency, Hz

Y1 amplitude, N·s^2/m

Y2 phase, rad

Figure E.1 — Selected acceleration to spine force transfer function (amplitude and phase)

The roughness of the curve in Figure E.1 for the low frequency area is caused by two related effects:

a) The seat-back and the typical seating postures of drivers are not parallel to the z-axis of the basicentric coordinate system defined in ISO 2631-1. Therefore, the excitation in z-axis does not only excite the vertical eigenmode of the model but also the horizontal (fore-and-aft) eigenmode which is in the area of 2 Hz.

b) Additionally, the forces calculated by the model are given within the local coordinate systems of the intervertebral discs (compression and shear). Due to the curvature of the lumbar spine and the seating posture, the inclination of some discs can be quite large. As a result, a fairly large compression force can be caused by the horizontal eigenmodes during vertical excitation.

The reason why this is not very obvious in the published measurement data, e.g. in seat-to-head transfer functions, is that the measurements are normally taken in the basicentric coordinate system and that the internal movements of the vertebra, which lead to the compression forces, cannot be measured. Additionally, during measurements at the head, subjects are able to partially suppress these movements

