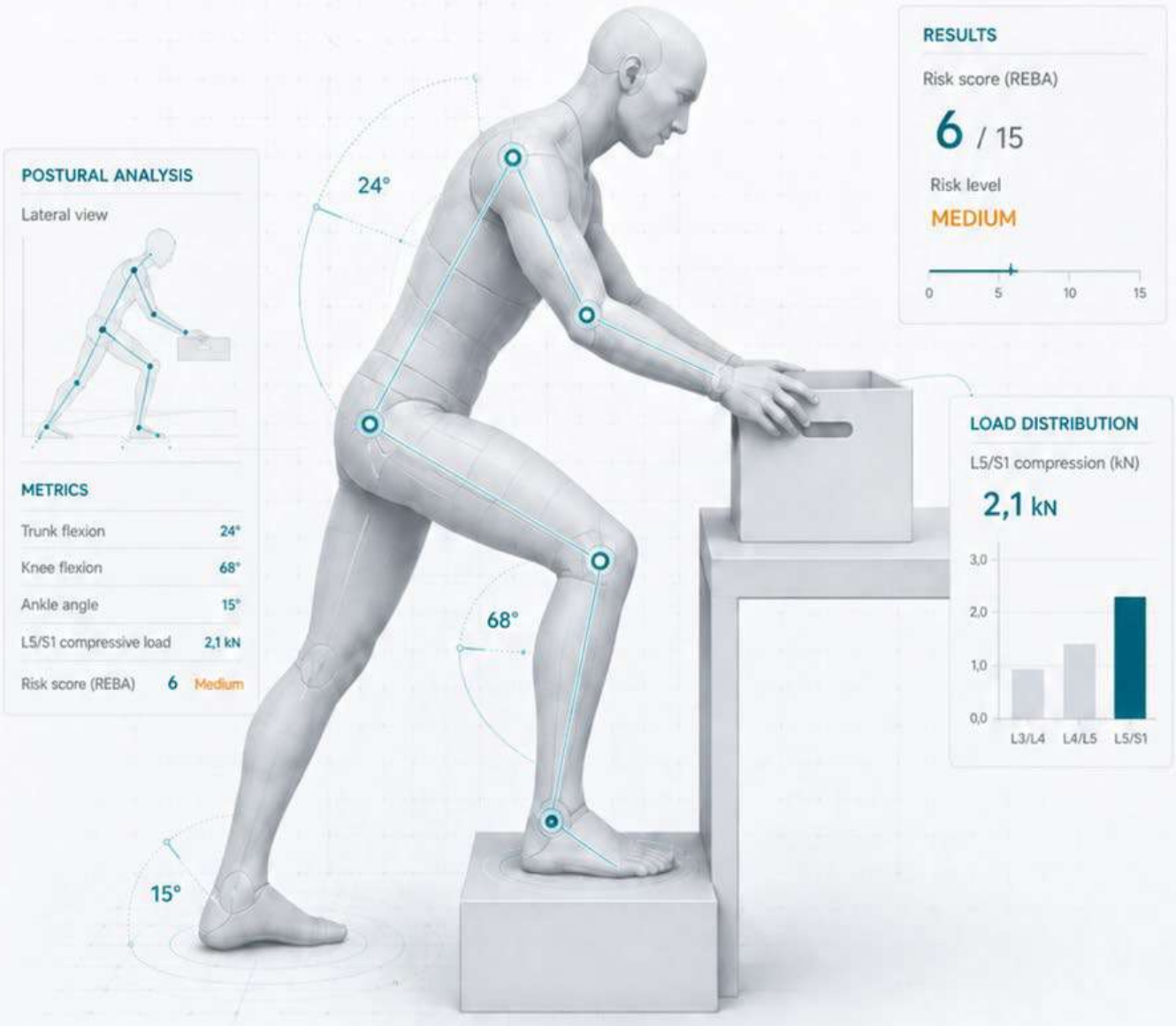
JOSÉ ANTONIO DIEGO MÁS

Ergoniza Intelligent

Technical foundations

Biomechanical and ergonomic assessment of postures, exertions, and movements using digital human models and artificial intelligence.



ergonautas

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Version 2026

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1 INTRODUCTION

Ergoniza Intelligent© is a support system for the ergonomic assessment and design of tasks, activities, and workstations involving human physical activity. Through an interactive graphical interface, it makes it possible to characterize posture and task conditions and obtain indicators related to biomechanical exertion, postural load, stability, and other risk factors described in this document.

In addition to identifying potential ergonomic risks, Ergoniza Intelligent can assess the effects of improvements or redesigns introduced into an activity by comparing the indicators obtained before and after their implementation. Its browser-based operation makes these capabilities readily accessible to professionals in ergonomics, occupational risk prevention, engineering, and design.

1.1 What is Ergoniza Intelligent?

Ergoniza Intelligent is a digital twin that provides a comprehensive analysis of how different tasks and activities affect the musculoskeletal system. Ergoniza Intelligent combines a biomechanical model with an interactive graphical interface and artificial-intelligence-powered systems for capturing the activity performed. To maximize accessibility, all calculation and data-capture processes are implemented in a web application that can be accessed over the Internet using a web browser such as Google Chrome (recommended), Microsoft Edge, or Safari. To protect data privacy, all calculations are performed locally on the user's device, and no information is sent to external servers.

1.1.1 What Ergoniza Intelligent is not

Ergoniza Intelligent is **not** an advanced musculoskeletal simulation system designed to reproduce the internal dynamics of the human body in maximum detail. Specialized tools are available for constructing highly detailed biomechanical models with specific representations of muscles, tendons, joints, muscle activations, internal forces, and complex dynamic parameters. Such systems are suitable for advanced biomechanical research, clinical simulation, detailed musculoskeletal analysis, or studies that require highly accurate modeling of human movement and the anatomical structures involved.

The purpose of Ergoniza Intelligent is different. It is not intended to provide an exhaustive experimental or clinical biomechanical analysis, but to provide useful, interpretable, and applicable information for the ergonomic assessment of real tasks. The system is designed especially for occupational risk prevention professionals, ergonomists, safety practitioners, engineers, and workstation designers who need to go beyond traditional observational methods without resorting to simulation tools that are complex, costly, or difficult to apply in typical workplace settings. Ergoniza Intelligent therefore seeks a balance among biomechanical foundations, ease of use, analysis speed, and practical interpretation of results.

1.2 What hardware do I need to use Ergoniza Intelligent?

Ergoniza Intelligent runs in a modern web browser and does not require the application to be installed on the device. An up-to-date version of Google Chrome is recommended because it provides the greatest compatibility. Other current browsers may also support the application, although some features may differ in compatibility or performance.

Performance depends on the device's processing power, memory, and graphics capabilities, particularly when analyzing long sequences or using artificial-intelligence-based capture features. Real-time motion capture requires a webcam or compatible image-capture device. An Internet connection is also required to access the application and use features that require remote processing.

1.3 What types of analysis can I perform with Ergoniza Intelligent?

Ergoniza Intelligent can perform different types of analysis and provide information about an activity and its potential effects on the person being assessed. Among other results, it can estimate exertions and joint loads, the percentage of maximum recommended exertion, the potentially unprotected population, cumulative exertion load, postural discomfort, REBA and LUBA scores, stability and the possibility of slipping, support reaction forces, and compression and shear loads in the lumbar region. In Lifting mode, it can calculate the Recommended Weight Limit and Lifting Index for a task, as well as the Composite Lifting Index when the activity contains several tasks. The temporal organization may comprise one or more lifting periods separated by recovery periods. For posture sequences, it can also analyze the evolution of different indicators and certain kinematic variables over time.

1.4 What data are required to perform an analysis with Ergoniza Intelligent?

Ergoniza Intelligent is a digital twin that models a person performing a particular activity. It therefore requires two types

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of information: information about the person performing the activity and information about the activity itself.

- The basic information about the person being assessed consists of sex, age range, height, weight, and body build. From these data, Ergoniza Intelligent calculates various biomechanical parameters, such as the length and weight of each body segment and the location of its center of gravity.
  Additional information can be provided for more advanced analyses, such as the coefficient of friction between the footwear and the floor, shoe-sole height, or the person's dominant side (right- or left-handed).

- The information required about the activity depends on the selected operating mode. In general terms, it may include the posture or sequence of postures adopted, external loads or forces, supports, and the corresponding temporal information.
  If the analysis focuses on a particular posture, the posture can be entered through the interactive graphical interface or captured from an image using the posture-detection systems. If the activity is analyzed over time, a posture sequence and its temporal information must be established. The sequence can be obtained from a video, imported from a motion-capture system, or captured in real time using a webcam or other compatible device. Lifting analysis also requires the lifting tasks and their temporal organization to be defined. Each task is characterized by its load, frequency, coupling, and origin and destination conditions. Independently, the temporal organization specifies the periods during which the task profile is performed and the recovery periods between them.

1.5 Responsibility for use

Should I use the information obtained from an Ergoniza Intelligent analysis as the sole criterion for determining an activity's risk level?

No. The results provided by Ergoniza Intelligent are estimates based on the data entered, the selected configuration, the implemented models, and their corresponding assumptions and limitations. They must be interpreted together with observation of the task, applicable methods and technical guidelines, and the judgment of a competent professional. Detections, assessments, or recommendations generated using artificial intelligence must be reviewed and verified before decisions are made or preventive measures are implemented. The assessor bears ultimate responsibility for interpreting and using the results.

Ergoniza Intelligent is particularly useful for occupational risk prevention, engineering, and design. However, the criteria applied and the decisions made must be based on the sound judgment of the professional using the tool, who bears ultimate responsibility for the decisions adopted.

2 ERGONIZA INTELLIGENT OPERATING MODES

Ergoniza Intelligent can analyze human physical activities in three main operating modes: **Single Posture**, **Multiple Postures**, and **Lifting**. Each mode addresses a different type of ergonomic problem and uses a different unit of analysis.

In **Single Posture** mode, the analysis focuses on a specific body configuration. The assessor can define or capture a posture, enter the load and support conditions, and obtain results related to biomechanical exertion, joint loads, stability, lumbar compression and shear loads, the REBA and LUBA methods, or postural discomfort. This mode is suitable for assessing a representative posture, a critical posture, or a particular work situation.

In **Multiple Postures** mode, the analysis is performed on a time sequence of postures that may represent a complete activity, a work cycle, or a significant part of a task. This mode accounts for posture duration, transitions between postures, the evolution of loads over time, cycle repetition, and recovery periods. It is therefore suitable for analyzing cumulative phenomena such as cumulative exertion load or the evolution of postural discomfort over time.

In **Lifting** mode, the assessment is organized into one or more time periods and one or more lifting tasks. Each task is characterized by variables such as load weight, frequency, coupling quality, and load location at the origin and destination. The temporal organization defines the duration of the lifting periods and the recovery periods between them. All periods currently use the same task and frequency profile. From these data, the system calculates the Recommended Weight Limit and Lifting Index for each task and the Composite Lifting Index when several tasks are analyzed, following the procedure described in Section 13.

The selected mode determines the data that must be entered, the calculations available, and how the results should be interpreted. Before beginning an assessment, select the mode that best matches the type of task to be analyzed.

The operating mode is selected using the buttons at the upper center of the interface, as shown in Figure 1.

When switching modes, some data, postures, or settings that are incompatible with the new mode may be deleted or reset.

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Ergoniza Intelligent displays a warning before making these changes. To preserve the current assessment, save it before confirming the mode change.



Figure 1: Selecting the operating mode in Ergoniza Intelligent.

3 UNITS, SYSTEMS, AND REFERENCE PLANES

Both the information required to perform an analysis with Ergoniza Intelligent and the resulting output use the International System of Units, apart from specific exceptions. Forces are expressed in newtons (N), distances in meters (m), and moments in newton-meters (N·m). Angles are generally displayed in degrees (°), and time may be expressed in seconds or minutes depending on the type of analysis.

To make certain data easier to enter and read, the interface also uses centimeters (cm) for some distances and kilograms (kg) for the mass of handled loads. When a mass entered in kilograms must be incorporated into the calculations as a weight force, Ergoniza Intelligent uses a standard gravitational acceleration of 9.81 m/s^2.

Ergoniza Intelligent uses two main reference systems to express position, forces, moments, and other vector quantities: the **Global Reference System** and the **Model Reference System**, shown in Figure 2.

Ergoniza Intelligent also uses multiple internal reference systems that are not displayed to the user. Many of them are used in calculations and are associated with the model's different body segments.

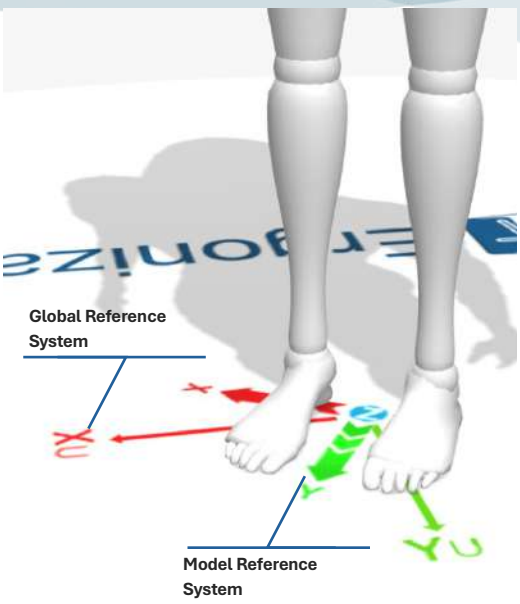


Figure 2: Global and model reference systems.

The reference-system axes are represented using the standard color convention: X in red, Y in green, and Z in blue.

3.1 Global reference system

The **Global Reference System** (Figure 2) is a fixed system associated with the scene. Its origin is located on the floor at the projection of the midpoint between the hips in the model's initial position. Its axes do not change orientation when the model is reoriented.

In the initial position, the ZG axis is vertical and positive upward; the XG axis is horizontal and positive toward the model's right; and the YG axis is perpendicular to the other two and positive toward the front of the model.

3.2 Model reference system

The **Model Reference System** is centered at the midpoint between the model's hips and is associated with its orientation. In the interface, however, it is shown projected onto the floor (Figure 2). The X axis extends horizontally toward the model's right, and the Z axis extends vertically upward. The Y axis is perpendicular to the other two and positive toward the front of the model.

When the model's global orientation changes, the Model Reference System changes with it and no longer coincides with the Global Reference System. A given force or moment retains its magnitude when expressed in either system, but its X, Y, and Z components may differ. The Model Reference System facilitates interpretation relative to the body, whereas the

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Global Reference System enables results to be interpreted relative to the scene's fixed axes. Use the same reference system when comparing results.

3.3 Principal reference planes and anthropometric planes

Three principal reference planes are defined from the axes of the Model Reference System, as shown in Figure 3. These planes are associated with the model's orientation and do not remain fixed relative to the scene.

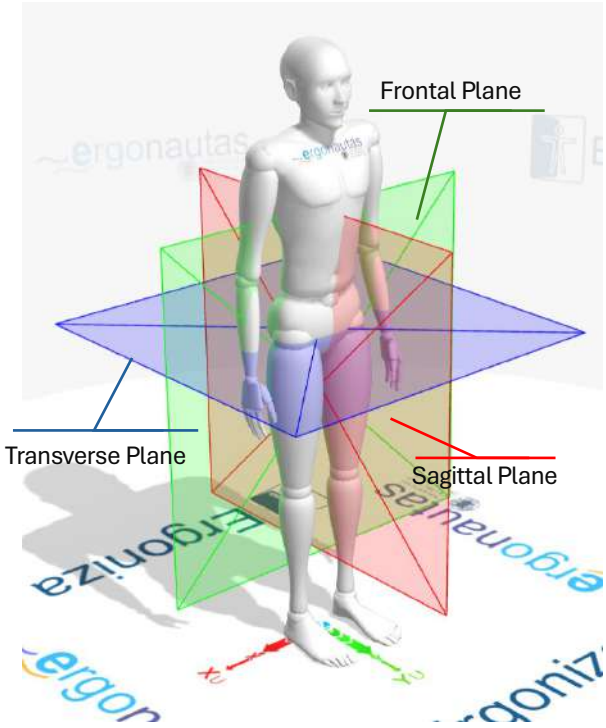


Figure 3: Anthropometric reference planes

- The **XY** plane contains the X and Y axes of the Model Reference System. This plane is horizontal in the model's initial position and coincides with the transverse anthropometric plane. Its positive normal vector is aligned with the Z axis of the Model Reference System and points upward.
- The **XZ** plane contains the X and Z axes of the Model Reference System. In the initial position, it is vertical and coincides with the frontal anthropometric plane. Its positive normal vector is aligned with the Y axis and points toward the front of the model.
- The **YZ** plane contains the Y and Z axes. In the initial position, it is vertical and coincides with the sagittal anthropometric plane. Its positive normal vector is aligned with the X axis and points toward the model's right.

These elements can be displayed using the Reference system, Sagittal plane, Frontal plane, and Transverse plane controls in the Visualization menu on the right side of the interface.

4 ANGLES BETWEEN BODY SEGMENTS

Ergoniza Intelligent describes posture numerically using the angles of the different body segments. To view these values, click **Angles** on the left side of the interface. The **Postural information** dialog opens (Figure 4).

The dialog's **Reference** selector provides three ways to present the angles: **Relative**, **Biomechanical**, and **Absolute**. Values are expressed in degrees. The selected reference does not change the model's posture; it only changes the criterion used to calculate and display the angles.



Figure 4: Postural information dialog

As a general rule, the **Relative** reference displays angles between connected body segments. Each segment is considered relative to the preceding segment in the kinematic chain. For example, upper-arm orientation is expressed relative to the torso, and forearm orientation is expressed relative to the upper arm.

The **Biomechanical** reference presents angles using the conventions applied in Ergoniza Intelligent biomechanical analyses. For the head, torso, and pelvis, it displays the corresponding flexion or extension, lateral flexion, and rotation movements. For the upper arms and thighs, it displays their vertical and horizontal orientation and their rotation about the longitudinal axis. For the remaining segments, it uses the biomechanical names corresponding to their movements.

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Calculating rotations, such as internal or external humeral rotation, is conceptually more complex than calculating other joint angles because it depends not only on segment direction but also on segment orientation about the longitudinal axis.

Although flexion or abduction can be estimated from the angle between axes, rotation requires a reference orientation to be defined and rotation to be measured about an axis that may become aligned with other body axes. The calculation of humeral and femoral rotation in Ergoniza Intelligent is explained in Appendix 15.1.

The **Absolute** reference displays segment orientation relative to the model's local reference system. The first two values correspond to the orientation of the segment's longitudinal axis relative to the **XZ** and **YZ** planes, whereas rotation is determined relative to the immediately preceding segment in the kinematic chain. In this context, *Absolute* means that the main orientation is not measured relative to the preceding segment; it should not be confused with the Global Reference System described in Section 3.1.

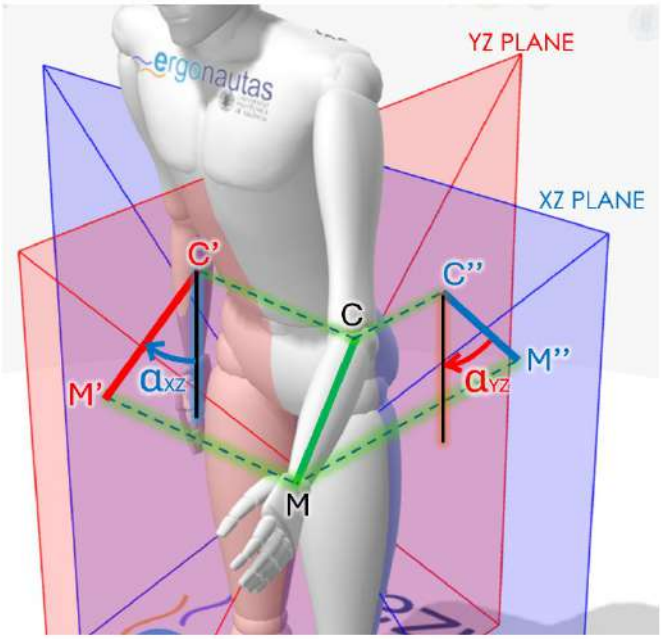


Figure 5: Calculating limb angles using the absolute reference system.

Figure 5 shows how these angles are calculated, using the left forearm as an example. Points C and M represent the elbow and wrist, respectively. To obtain the angle relative to the **XZ** plane, the C–M forearm axis is projected onto the **YZ** plane, producing the C'–M' projection. The α_{XZ} angle is measured in that plane between the projection and a line parallel to the **Z** axis, in the direction shown in the figure. Similarly, the α_{YZ} angle is obtained by projecting the C–M axis onto the **XZ** plane (C''–M'' projection) and measuring its angle relative to a line parallel to the **Z** axis.

5 THE ERGONIZA INTELLIGENT BIOMECHANICAL MODEL

Ergoniza Intelligent implements a full-body computational biomechanical model. The model represents the body as segments connected by joints and can estimate forces, moments, and other mechanical quantities generated by posture, anthropometric characteristics, external loads, and support conditions.

Biomechanical models can have different levels of complexity depending on their purpose. The Ergoniza Intelligent model considers the whole body and incorporates segments, joints, inertial parameters, force systems, and, in certain calculations, muscle models. Its results support ergonomic assessment but do not, by themselves, predict injury or harm to health.

The following sections describe the structure, assumptions, and fundamental parameters of the Ergoniza Intelligent biomechanical model.

5.1 Segments and joints

The biomechanical model implemented in Ergoniza Intelligent divides the entire body into 16 main segments (shown in black in Figure 6). In this count, the head and neck constitute two distinct subsegments in the kinematic chain, although certain biomechanical calculations treat them together as a single functional head-neck segment.

These segments are supplemented by two segments in each thumb, three in each of the other fingers, and two representing the toes of each foot. These additional segments form part of the articulated graphical representation but are not involved in every biomechanical calculation.

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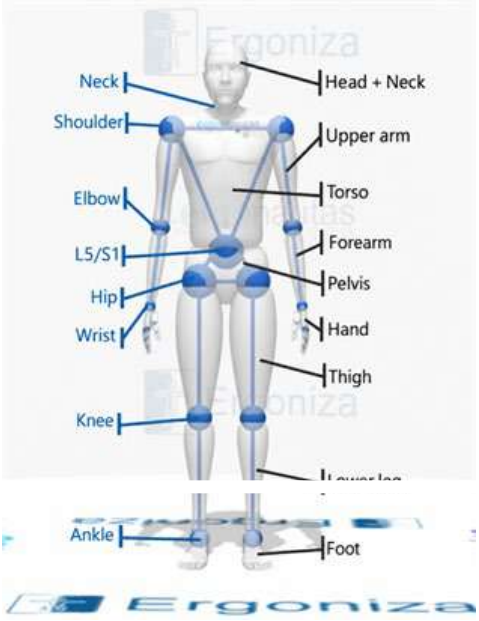


Figure 6: Segments and joints of the biomechanical model.

Figure 6 also shows fourteen main joints, represented in blue. Some joint levels used in specific calculations are not shown separately in the figure, such as the joint between the C1 vertebra and the skull or the L4/L5 intervertebral level.

In general, each segment is bounded by two joints. Exceptions are the pelvis, which connects the two hips to the L5/S1 lumbosacral joint, and the torso, which connects L5/S1 to the shoulders and the C7/T1 region.

In the model's neutral position, the joint centers of the hips, knees, and ankles lie in the frontal plane. The L5/S1 joint is located slightly behind this plane because of the configuration adopted for the lumbosacral joint, described in Section 5.1.2.1. Consequently, the joints located above L5/S1 also lie slightly behind the frontal plane.

The segments are considered rigid, and the joints are modeled as points located at their respective joint centers. Each segment's longitudinal axis begins and ends at a joint. The center of gravity of each segment is considered to lie on its longitudinal axis, except for the foot.

5.1.1 The head-neck segment

The head and neck are represented by two subsegments joined at the C1 vertebra. The neck subsegment extends from the C7/T1 region to C1, while the head subsegment extends from C1 to the vertex. For biomechanical calculations, they are considered together as a single functional head-neck segment.

The head-neck assembly has three degrees of freedom: flexion and extension, left and right lateral flexion, and left and right rotation. When any of these angles is changed, Ergoniza Intelligent distributes the total rotation equally between the neck and head subsegments.

5.1.2 The torso and pelvis segments

Ergoniza Intelligent models the trunk using two segments: the torso and the pelvis. The torso segment comprises the thorax and abdomen and extends from the L5/S1 lumbosacral joint to the C7/T1 region. The pelvis segment extends from L5/S1 to the hip joints.

The trunk has three degrees of freedom: flexion and extension, left and right lateral flexion, and left and right rotation. Lateral-flexion and rotation movements are performed by the torso segment through its rotation relative to the pelvis at L5/S1, while the pelvis remains fixed. In contrast, trunk flexion and extension are distributed between the torso and pelvis segments. This distribution is explained in the following section.

5.1.2.1 Modeling flexion at the lumbosacral joint

In the Ergoniza Intelligent model, total trunk flexion results from the combination of two movements: rotation of the pelvis relative to the thighs at the hip joints and flexion of the torso relative to the pelvis in the lumbosacral region. Several studies have shown that the thoracic vertebrae exhibit little relative movement during trunk flexion and that a substantial part of the movement occurs in the lumbar region [1].

Flexion from an upright position begins with rotation of the lumbar vertebrae. In general, as trunk flexion increases, the pelvis begins to rotate together with the spine, although at a different rate. The rotation rates assumed by the spine and pelvis have been analyzed in different studies to obtain a

[1] Peter R. Davis, "Posture of the Trunk during the Lifting of Weights," *British Medical Journal*, 1959, doi:10.1136/bmj.1.5114.87; S. Kumar, "A Study of Spinal Motion during Lifting," *Irish Journal of Medical Science* 143, no. 2 (1974): 86–95, doi:10.1007/BF03004745; Elizabeth A. Tully, Mohammad Reza Fotoohabadi, and Mary P. Galea, "Sagittal Spine and Lower Limb Movement during Sit-to-Stand in Healthy Young Subjects," *Gait and Posture*, 2005, doi:10.1016/j.gaitpost.2004.11.007; P. R. Davis, J. D. Troup, and J. H. Burnard, "Movements of the Thoracic and Lumbar Spine When Lifting: A Chrono-Cyclophotographic Study," *Journal of Anatomy* 99 (1965): 13–26.

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quantitative model of spinal kinematics during trunk flexion[2]. These studies have shown that the degree of sacral rotation at the hip joints and the rotation of the L5 vertebra relative to the sacrum depend not only on the degree of trunk flexion but also on the knee-flexion angle. Other factors, such as the load handled in the hands, sex, body weight, or somatotype, appear to have a smaller effect on the lumbosacral-joint rotation pattern[3].

Ergoniza Intelligent models the relative movements between the torso and pelvis segments during trunk flexion and extension using these kinematic models. At the beginning of flexion from an upright position, most rotation occurs in the lumbar region. As the trunk-flexion angle increases, the rate of sacral rotation at the hips increases until it predominates over rotation in the lumbar region. The effect of knee flexion is also considered: as knee flexion increases, the relative amount of sacral rotation decreases, while relative rotation at L5/S1 increases.

An example is shown in Figure 7. The postures in the figure have the same trunk flexion, while knee flexion is greater in posture B. The distribution of flexion between the pelvis and torso differs between the two postures. Sacral rotation is lower in posture B because knee flexion is greater.

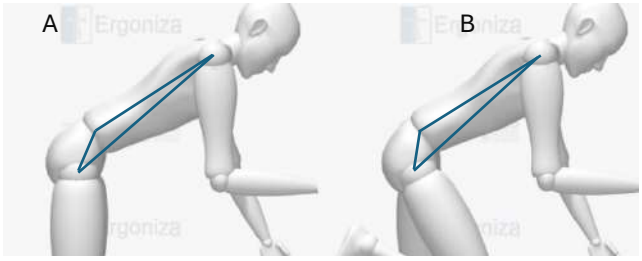


Figure 7: Variation in the relative rotation rate between the pelvis and torso with the degree of knee flexion.

5.1.2.2 Modeling the trunk musculature

Ergoniza Intelligent estimates compression and shear forces at the L4/L5 intervertebral disc. The magnitude of these forces and their comparison with the corresponding reference values provide relevant information for assessing tasks that may generate high loads in the lumbar spine. [4]

To solve moment equilibrium at L4/L5, the model incorporates ten muscle units distributed across five bilateral pairs: *latissimus dorsi*, *erector spinae*, *rectus abdominis*, *obliquus externus*, and *obliquus internus*. For each unit, the model considers its line of action, its moment arm relative to the spine, and its cross-sectional area at L4/L5[5]. The calculation also incorporates the effect of intra-abdominal pressure.

The procedure used to determine muscle forces and calculate compression and shear forces at L4/L5 is described in detail in Section 12.

5.1.3 The upper arm and thigh segments

The upper-arm segment extends from the shoulder to the elbow and has three degrees of freedom: abduction/adduction, anterior/posterior rotation, and humeral rotation. The thigh extends from the hip to the knee and also has three degrees of freedom: abduction/adduction, anterior/posterior rotation, and femoral rotation.

5.1.4 The forearm and lower leg segments

The forearm segment extends from the elbow to the wrist, while the lower-leg segment extends from the knee to the ankle. In the biomechanical model, both segments have a single degree of freedom: flexion/extension. Axial rotations of the forearm and lower leg are not assigned to these segments. Instead, they are represented in the next segment of the kinematic chain: by pronation/supination of the hand segment and abduction/adduction of the foot segment, respectively.

5.1.5 The hand segment

In the biomechanical model, the hand segment extends from the wrist to the distal end of the hand and has three degrees of freedom: flexion/extension, radial/ulnar deviation, and pronation/supination. Although pronation/supination occurs anatomically in the forearm, Ergoniza Intelligent represents this rotation by rotating the hand segment at the wrist joint.

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[2] C. K. Anderson, D. B. Chaffin, and G. D. Herrin, "A Study of Lumbosacral Orientation under Varied Static Loads," *Spine* 11, no. 5 (1986): 456–62, doi:10.1097/00007632-198606000-00012; C K. Anderson et al., "A Biomechanical Model of the Lumbosacral Joint during Lifting Activities," *Journal of Biomechanics* 18, no. 8 (1985): 571–84, doi:10.1016/0021-9290(85)90012-0.

[3] D. Allbrook, "Movements of the Lumbar Spinal Column.," *The Journal of Bone and Joint Surgery. British Volume*, 1957, doi:10.1302/0301-620x.39b2.339.

[4] T R Waters et al., "Revised NIOSH Equation for the Design and Evaluation of Manual Lifting Tasks.," *Ergonomics* 36 (1993): 749–76, doi:10.1080/00140139308967940.

[5] S. McGill, *Low Back Disorders. Evidence-Based Prevention and Rehabilitation* (Champaign, Illinois: Human Kinetics, 2007); D. B. Chaffin et al., "Lumbar Muscle Size and Locations from CT Scans of 96 Women of Age 40 to 63 Years," *Clinical Biomechanics* 5, no. 1 (1990): 9–16, doi:10.1016/0268-0033(90)90026-3.

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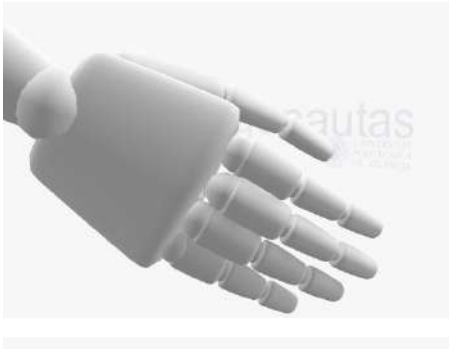


Figure 8: Distal, middle, and proximal phalanges at the end of the hand segment.

As shown in Figure 8, the graphical model of Ergoniza Intelligent incorporates articulated segments representing the distal, middle, and proximal phalanges of each finger. These elements serve a graphical function only: they are not considered in the biomechanical model or included in the analyses.

5.1.6 The foot segment

In the biomechanical model, the foot segment extends from the ankle to the distal end of the foot and has three degrees of freedom: flexion/extension, abduction/adduction, and inversion/eversion. Axial rotation of the lower leg that is not assigned to the lower-leg segment is represented by abduction/adduction of the foot segment. The graphical model also incorporates an articulated segment that represents the distal, middle, and proximal phalanges of each foot together. This element serves a graphical function only and is not considered in the biomechanical model or the analyses.

5.2 Range-of-motion limits

Ergoniza Intelligent lets you define the posture adopted by the biomechanical model by directly manipulating the different segments of the graphical model in the user interface.

Human joints cannot rotate without limit. Therefore, Ergoniza Intelligent establishes a range between a minimum and a maximum angle for each available degree of freedom.

By default, **Apply biological constraints to the movement of the model's joints**, located under **Model settings** on the right side of the interface, prevents direct manipulation from exceeding these limits. For certain joints, it also applies geometric constraints intended to prevent incompatible configurations between segments. If this option is disabled, the available degrees of freedom may exceed the configured angular limits; axes that the model does not include as degrees of freedom remain locked.

Disabling the constraints may produce anatomically implausible or impossible postures. Results obtained from such postures may fall outside the model's validity range and must be interpreted with particular caution.

Joint ranges of motion can vary substantially among individuals because of factors such as age, sex, body build, injury history, and physical activity. For this reason, the limits implemented in Ergoniza Intelligent are broad. They should be interpreted as operational software limits, not as normative population values or as a guarantee that every posture within those limits is suitable for a particular person.

Segment	Movement	Maximum	Minimum
Pelvis	Flexion/extension	80º	-7º
Torso	Flexion/extension	140º	-30º
	Left/right rotation	60º	-60º
	Left/right lateral flexion	35º	-35º
Head-neck	Flexion/extension	90º	-50º
	Left/right rotation	90º	-90º
	Left/right lateral flexion	45º	-45º
Upper arm	Flexion/extension	198º	-49º
	Abduction/adduction	90º	-26º
	Humeral rotation	90º	-90º
Forearm	Flexion/extension	165º	0º
Hand	Flexion/extension	90º	-90º
	Radial/ulnar deviation	42º	-40º
	Supination/pronation	90º	-90º
Thigh	Flexion/extension	130º	-40º
	Abduction/adduction	60º	-30º
	Femoral rotation	70º	-70º
Knee	Flexion/extension	160º	0º
Foot	Flexion/extension	70º	-80º
	Abduction/adduction	30º	-30º
	Inversion/eversion	40º	-35º

Table 1: Range-of-motion limits by segment

The range-of-motion limits for each segment are provided in Table 1. For the upper-arm and thigh segments, flexion/extension describes movement in a plane approximately parallel to the sagittal plane, whereas abduction/adduction describes movement in a plane approximately parallel to the frontal plane.

5.3 Inertial parameters

Ergoniza Intelligent estimates the mass, length, and center-of-gravity location of each body segment from the sex, height, and body mass entered for the person being assessed. In the interface, the estimated mass of each segment is shown under **Weight** and expressed in kilograms. These estimates use

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segmental anthropometric parameters obtained from studies of living people and cadavers [6].

To estimate lengths, Ergoniza Intelligent uses anthropometric relationships that express the length of each segment as a proportion of height. Drillis and Contini obtained these relationships from measurements of living subjects and performed a statistical regression against the *height* variable. This makes it possible to estimate the length of each segment as a proportion of the individual's height.

Segment masses and center-of-gravity locations are estimated using coefficients that relate these quantities to total body mass and the length of each segment. Because these are population-based estimates, the resulting values should not be interpreted as direct measurements of the person being assessed.

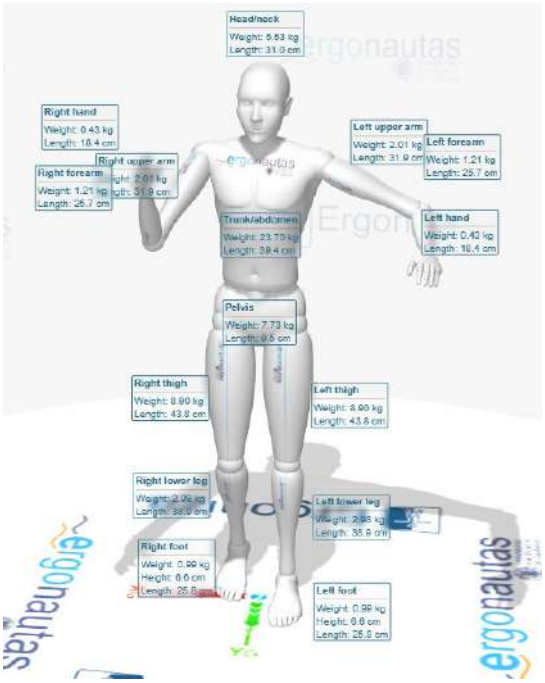


Figure 9: Estimation of body-segment mass and length.

Figure 9 shows the estimated mass and length of the body segments for a person with a body mass of 70 kg and a height of 1.70 m. To display these data, enable **Dimensions** in the **Visualization** panel on the right side of the interface.

5.3.1 Centers of gravity

Once the model's posture and the mass and center-of-gravity location of each segment are known (the segmental centers of gravity shown in Figure 10), Ergoniza Intelligent calculates the position of the body's center of gravity using the segmental method [7]. The resulting position is obtained as the weighted average of the segmental center-of-gravity positions, so that each segment's contribution is proportional to its mass.

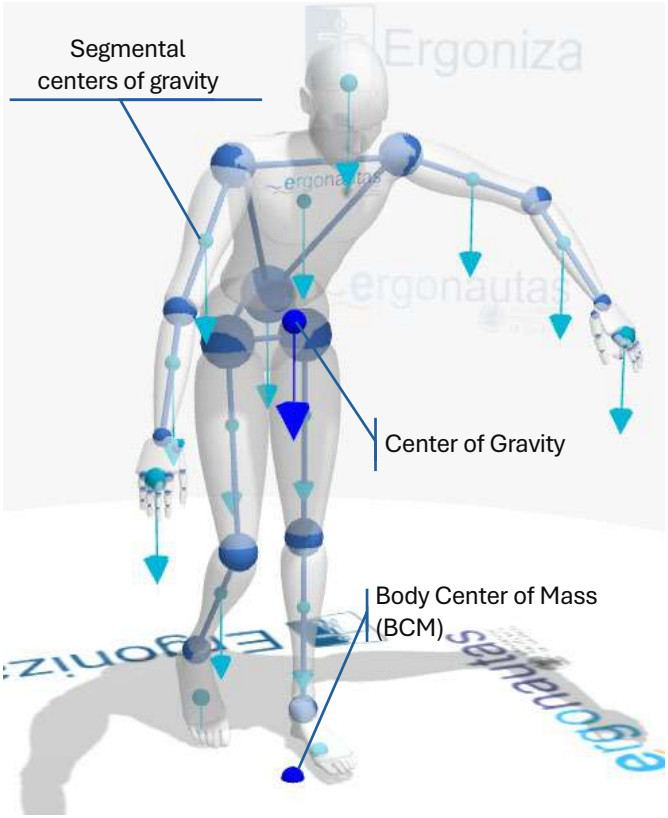


Figure 10: Calculating the body's center of gravity from segmental centers of gravity.

External loads and moments applied at the hands are not included in the calculation of the body's center of gravity. Their effect is incorporated into the center-of-pressure calculation described in Section 6. Forces applied to other parts of the body are, however, included in the calculation, taking into account the segment and the point on which they act.

[6] W. T. Dempster, *Space Requirements of the Seated Operator. (Technical Report WADC-TR-55-159), Wright-Patterson Air Force Base, OH: Aerospace Medical Research Laboratory* (Dayton, 1955); Charles E Clauser, John T McConville, and J W Young, "Weight, Volume, and Center of Mass of Segments of the Human Body," *National Technical Information Service*, 1969; R. Contini, R. J. Drillis, and M. Bluestein, "Determination of Body Segment Parameters," *Human Factors: The Journal of Human Factors and Ergonomics Society* 5, no. 5 (1963): 493–504, doi:10.1177/001872086300500508; P. De Leva, "Adjustments to Zatsiorsky-Seluyanov's Segment Inertia Parameters," *Journal of Biomechanics* 29, no. 9 (1996): 1223–30, doi:10.1016/0021-9290(95)00178-6.

[7] D. G. E. Robertson et al., *Research Methods in Biomechanics*, 2nd Ed. (Human Kinetics Publishers, 2014), doi:10.5040/9781492595809.

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Under the conditions considered, the body's center of mass and center of gravity are treated as coincident. To analyze stability, Ergoniza Intelligent represents their vertical projection onto the floor, identified in the interface as the **Body Center of Mass** (BCM).

6 BASE OF SUPPORT, CENTER OF PRESSURE, AND STABILITY

The **center of pressure (CoP)** is the point of application of the resultant reaction forces exerted by the floor on the supports. Ergoniza Intelligent calculates its position on the floor plane from force and moment equilibrium, accounting for body weight, external loads or forces applied at the hands, and external moments applied at them. When none of these loads, forces, or moments acts, the CoP coincides with the vertical projection of the **body center of mass (BCM)**. Figure 11 shows a situation in which the two positions do not coincide because of a load held in the hands.

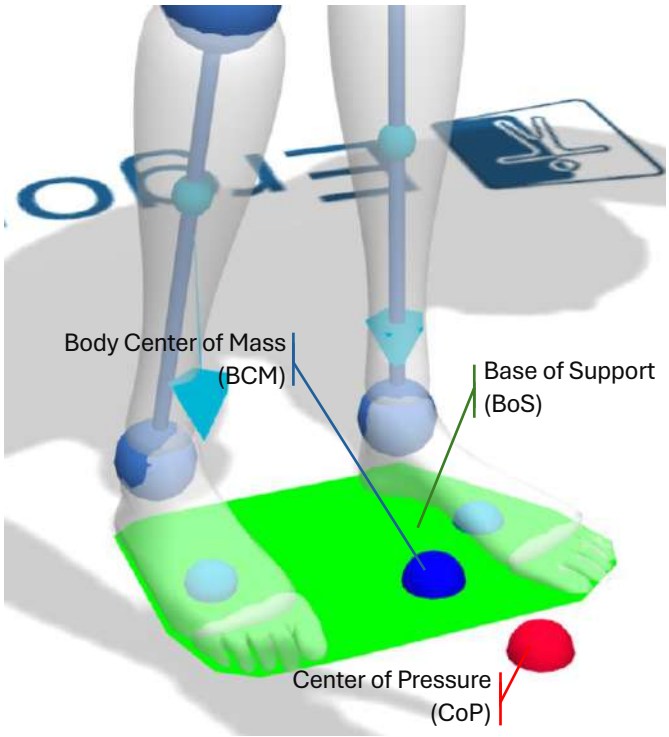


Figure 11: Location of the center of pressure on the base of support

For a posture to remain in static equilibrium, the forces and moments generated at the supports must balance those produced by the weight of the body segments, external loads or forces, and applied moments. In the model, this requires the center of pressure to remain within the **base of support (BoS)**. The base of support is bounded by the effective contact points between the body and the floor; during normal bipedal stance, it corresponds to the area defined by the two foot supports.

In the case shown in Figure 11, the center of pressure lies outside and in front of the base of support, so the model identifies instability with possible forward tipping. The body center of mass, however, remains within the base of support. Therefore, in the absence of external loads, forces, or moments applied at the hands, the represented posture would be stable.

To display the segmental centers of gravity, body center of mass, center of pressure, and base of support, use the **Stabilogram** (Figure 30), described in Section 7.8.

6.1 Global slipping

When external loads or forces include horizontal components, maintaining the posture without slipping requires the horizontal floor reaction to balance them. Static friction can provide this reaction up to a certain maximum value. Therefore, to prevent slipping, the magnitude of the horizontal resultant that must be supported by the contact cannot exceed the maximum available static-friction force, as shown in Figure 12.

Ergoniza Intelligent calculates this maximum force using the Coulomb friction model:

$$F_{fric,max} = \mu \cdot F_{normal}$$

Equation 1

where $F_{fric,max}$ is the maximum available static-friction force, F_{normal} is the magnitude of the total normal reaction at the supports, and μ is the coefficient of static friction between the footwear and the floor. The normal reaction depends on body weight and the vertical components of external forces, accounting for their direction. The coefficient of friction is set under **Model settings**, on the right side of the interface, using the **Coefficient of friction** control.

When the preceding criterion indicates slipping, static equilibrium cannot be maintained with the selected coefficient of friction. Slipping may occur before the supports develop the reactions required to cause tipping. Nevertheless, Ergoniza Intelligent assesses and displays the two conditions separately: the possibility of slipping and the location of the center of pressure relative to the base of support. This lets the user determine whether, even if slipping were prevented—for example, by using footwear or a surface with a higher coefficient of friction—the posture would remain unstable because of tipping.

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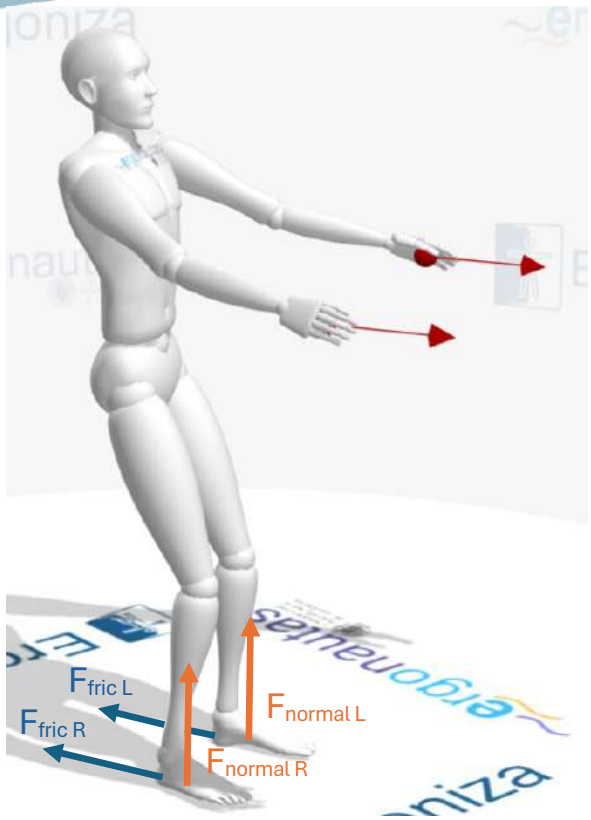


Figure 12: Calculation of slipping on the floor

The preceding criterion considers the overall friction capacity of the supports. During bipedal stance, however, the horizontal and vertical reaction forces are distributed between both feet. Even if the total friction capacity is sufficient, the horizontal reaction required at one support may exceed its individual capacity. Ergoniza Intelligent also checks this condition and, where possible, redistributes the reactions between the supports. The procedure and its interpretation are explained in Section 7.3, **Partial slipping**.

7 SUPPORT MODELING

The reaction at the contact between the body and the floor—generally at the soles of the feet—is a force distributed over the entire contact surface. If external forces or loads have no horizontal components, the contact forces needed to balance them will likewise have no horizontal components and will be normal to the floor. In general, the distribution of the reactions depends on the position of the center of pressure relative to the base of support and on the geometry of the active supports.

Ergoniza Intelligent can account for different contact areas between the body and the floor. Specifically, it detects the soles of the feet, the front of the knees, and two points on the pelvis located approximately at the ischial tuberosities.

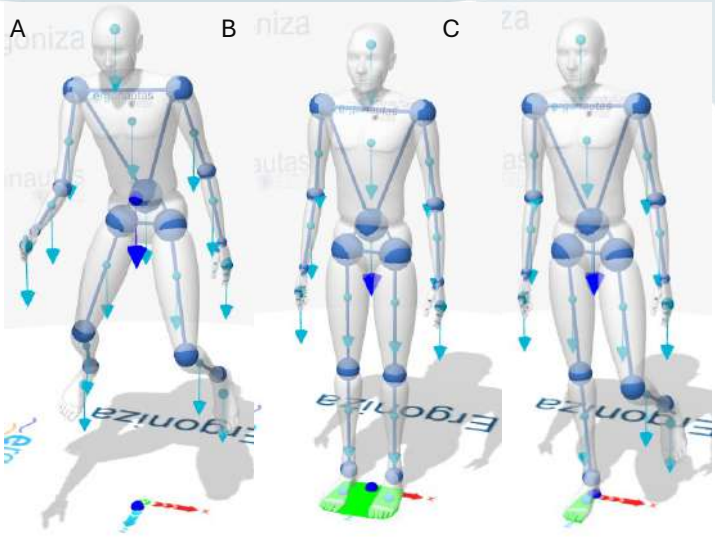


Figure 13: Type of contact between the model and the floor; A: No support, B: Bipedal stance, C: Single-leg stance

Although the actual reaction is distributed over a surface, Ergoniza Intelligent represents it by means of forces concentrated at a limited number of points. Each foot has two possible support points: the heel at the rear and the central metatarsal contact area at the front. Contacts at the knees and ischial tuberosities are also represented as points.

A linear variation in pressure is assumed when distributing the reactions among these points. In addition, the vertical components of the reactions can only act upward: a support can separate from the floor, but it cannot exert a downward tensile force. This condition represents the unilateral nature of contact and prevents negative reactions from being used to stabilize a posture whose center of pressure lies outside the base of support. It implies, for example, that the footwear is not rigidly attached to the floor and therefore cannot generate moments that counteract an unbalanced posture.

The calculation for contacts at the feet is explained first. The procedure is then adapted to knee supports in Section 7.5, pelvic supports in Section 7.6, and seated posture on a chair in Section 7.7.

Considering foot contacts first, Ergoniza Intelligent distinguishes three situations, shown in Figure 13: **no support**, when neither foot produces an effective reaction against the floor; **bipedal stance**, when both feet produce reaction forces; and **single-leg stance**, when only one foot does so.

Detecting physical proximity between a foot and the floor is not sufficient to determine the support condition. An effective reaction force must also exist. Therefore, if both feet are in contact with the floor but one of them supports no reaction, the model is considered to be in single-leg stance.

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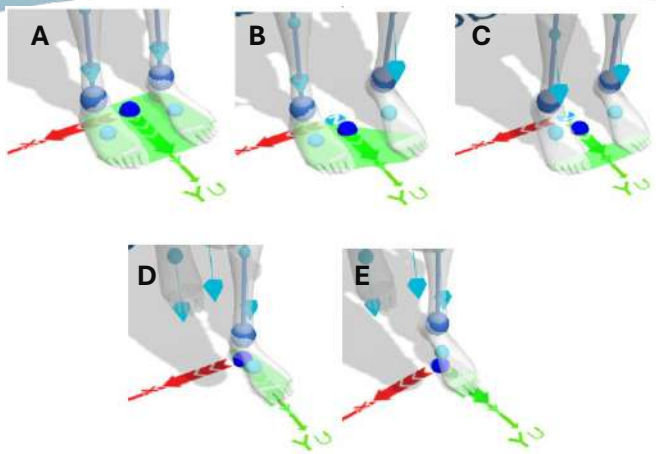


Figure 14: Different numbers of supports in bipedal and single-leg stance.

Because two possible support points are considered for each foot, bipedal stance can have between two and four active points (cases A, B, and C in Figure 14), whereas single-leg stance can have one or two (cases D and E).

7.1 Reaction forces at the foot supports during bipedal stance

During bipedal stance, Ergoniza Intelligent considers between two and four active contact points between the feet and the floor. The number depends on whether effective contact exists at the heel, the metatarsal area, or both parts of each foot.

For static equilibrium to exist, the resultant of the reactions at the supports must be equal and opposite to the resultant of body weight and the external forces. In addition, the sum of the moments produced by all forces and external moments must be zero. The position of the center of pressure already incorporates the effect of these external actions; therefore, the resultant of the vertical reaction components must pass through that point. When no external loads, forces, or moments are applied at the hands, the center of pressure coincides with the vertical projection of the body center of mass.

When four contact points exist, the equilibrium equations alone do not determine a unique distribution of the reactions: four vertical components remain unknown, but there are only three independent equilibrium conditions. Ergoniza Intelligent completes the system using the geometric criterion described in the following sections. When fewer contact points exist, the procedure is adapted to the available supports.

7.1.1 Calculation of the vertical components of the reaction forces at the foot supports

The calculation begins by determining the vertical components of the reaction forces. The sum of these components must equal the total normal reaction required to balance body weight and the vertical components of the external forces. In addition, their combined line of action must pass through the center of pressure so that their resultant moment about this point is zero.

Given these conditions, Ergoniza Intelligent calculates the reactions by distributing them between both feet and among the support points of each foot, based on the position of the ankles relative to the center of pressure (CoP).

The procedure is illustrated in Figure 15, which shows a bipedal stance with the red sphere indicating the position of the CoP on the base of support. Points T_d and P_d represent the heel and metatarsal area of the right foot, respectively; T_i and P_i represent the equivalent points on the left foot. Points X_d and X_i are the projections on the floor of the right and left ankle joint centers, respectively.

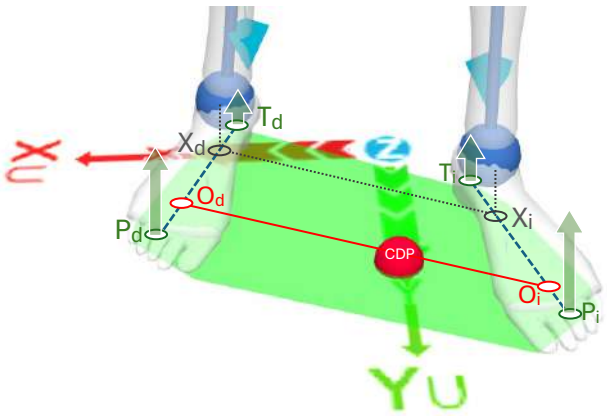


Figure 15: Vertical components of the reaction forces at the supports.

Vertical reaction forces act at the four contact areas: T_d at the right heel, P_d at the metatarsal area of the right foot, T_i at the left heel, and P_i at the metatarsal area of the left foot.

The line joining X_d and X_i is translated parallel to itself until it passes through the center of pressure. Its intersections with the lines joining the heel and metatarsal area of each foot determine the zero-moment points O_d and O_i. The vertical reaction is distributed between the heel and front of each foot in inverse proportion to the distance from each support point to its corresponding zero-moment point.

Once O_d and O_i have been obtained, the total reaction is distributed between both feet using the same equilibrium principle: the load supported by each foot is inversely

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proportional to the distance from the center of pressure to the corresponding zero-moment point. This geometric criterion, together with the equilibrium equations, yields a unique distribution of the vertical components.

Under certain circumstances—for example, when one foot is markedly rotated and the CoP is close to the boundary of the base of support—the zero-moment point may lie outside the segment between the heel and the front of the foot.

This situation is illustrated in Figure 16, where the zero-moment point of the right foot (O_d) lies outside the segment joining the foot contact points (P_d-T_d). The unconstrained solution would then require a negative vertical reaction at one of the points, which is equivalent to assuming that the foot can exert a tensile force on the floor.

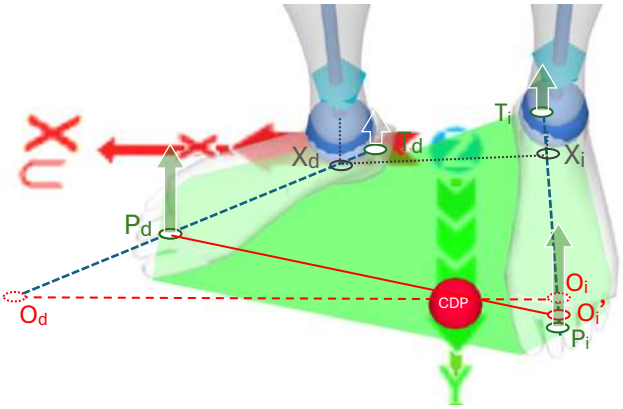


Figure 16: Compensation of the vertical support forces when the zero-moment point lies outside the foot.

Because negative vertical reactions are not admissible, Ergoniza Intelligent places the zero-moment point of the affected foot at the nearest support point. In the example, O_d is moved to P_d, the reaction at T_d becomes zero, and the entire reaction of the right foot is concentrated at P_d. In other words, placing the zero-moment point of the right foot at the front support causes the entire load on that foot to be carried by that support.

The line joining this point to the center of pressure determines a new zero-moment point O_i' on the left foot, from which the distribution between the front and rear supports of that foot is recalculated.

Furthermore, changing the distances between the CoP and the new zero-moment points increases the loads on the left foot—which is better positioned to support them—and therefore decreases those supported by the right foot.

A similar procedure is used in bipedal stance when one foot has only one contact area (Figure 17). The reaction of that foot is concentrated at this point. The line joining the contact to the CoP determines the zero-moment point (O_i) on the foot that retains two supports. The vertical reaction components are distributed between the two supports of this foot in inverse proportion to the distance from each support to the zero-moment point.

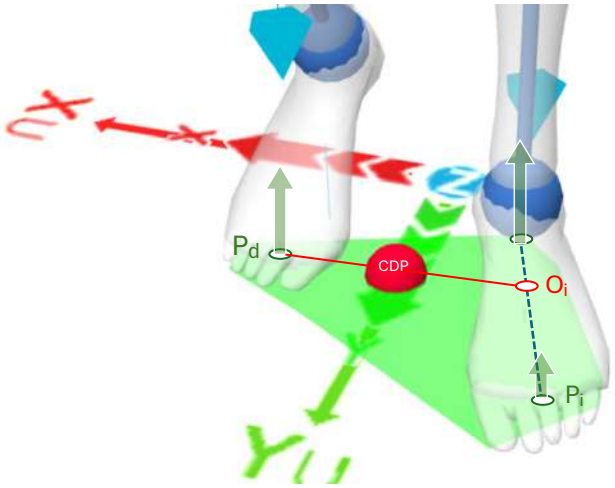


Figure 17: Vertical components of the reaction forces at the supports during bipedal stance with three support points.

7.1.1.1 Vertical components of the reaction forces at the foot supports under imbalance

When the center of pressure (CoP) lies outside the base of support (BoS), static equilibrium cannot be achieved using only nonnegative vertical reactions at the available supports. Ergoniza Intelligent then identifies the posture as unstable, but continues to calculate and display a diagnostic distribution of the reactions to indicate the edge or support toward which the load is concentrated and the direction of the tipping tendency.

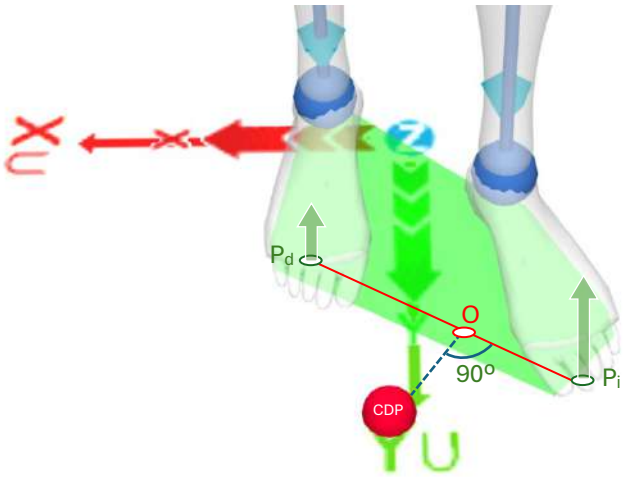


Figure 18: Vertical components of the reaction forces at the supports during forward imbalance in bipedal stance.

If the center of pressure lies in front of the base of support, as in Figure 18, the heel reactions are taken as zero and the entire

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vertical reaction is concentrated at the fronts of the feet. The distribution between the right and left feet is inversely proportional to the distances from each support to point O, defined as the perpendicular projection of the center of pressure onto the line joining both supports. If the center of pressure lies behind the base of support, the procedure is analogous, but the reactions are concentrated at the heels.

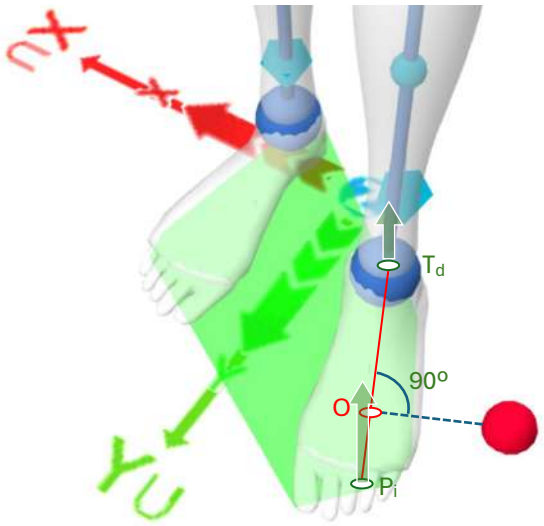


Figure 19: Vertical components of the reaction forces at the supports during lateral imbalance in bipedal stance.

If the center of pressure lies to one side of the base of support, the reactions are concentrated on the nearest foot. Figure 19 shows the case in which the center of pressure lies to the left: the reactions of the right foot are zero and the entire load is supported by the left foot. The distribution between the heel and the front of this foot is calculated from the perpendicular projection of the center of pressure onto the line joining the two support points. The procedure is symmetrical when the center of pressure lies to the right.

When the center of pressure lies outside the base of support near a corner, both criteria are combined and the reaction may be concentrated at the nearest support point.

7.1.2 Calculation of the horizontal components of the reaction forces at the foot supports during bipedal stance

Reaction forces have horizontal components when the applied external forces also have them. To satisfy static equilibrium, the sum of the horizontal reaction components must be equal and opposite to the horizontal resultant of the external forces. In addition, the moment that these forces produce about the Z axis passing through the center of pressure must be balanced by the moment generated by the reactions.

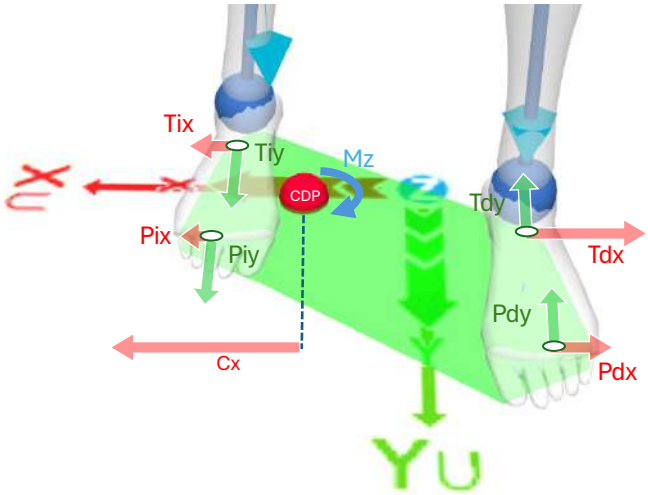


Figure 20: Distribution of horizontal reactions for X components of external loads.

These conditions alone do not determine a unique distribution between the feet, heels, and metatarsal areas. Therefore, Ergoniza Intelligent analyzes the components corresponding to the X and Y axes separately and introduces additional criteria related to the relative positions of the feet.

The following expressions and diagrams correspond to the case in which all four support points are active. If a foot remains in contact only at the heel or at the front, the components corresponding to that foot are concentrated at that single point.

Figure 20 shows the case in which the external forces have a component Cx. This component generates a moment Mz about the center of pressure, which must be balanced by the reaction components Tdx, Pdx, Tix, and Pix.

When the ankles have the same Y coordinate, the total reaction in X is distributed between both feet in proportion to the vertical reaction supported by each one. For example, if the right foot supports 70% of the total vertical reaction, the magnitudes satisfy:

$$\left|T_{d,x} + P_{d,x}\right| = 0.7|C_x|$$

$$\left|T_{i,x} + P_{i,x}\right| = 0.3|C_x|$$

These reactions act in the direction opposite to Cx. Within each foot, the reaction is distributed between the heel and the front so that the moment about the ankle is zero. The Y components form a pair of opposite reactions between both feet and are distributed equally between their support points:

$$T_{i,y} = P_{i,y} = -T_{d,y} = -P_{d,y}$$

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When the ankles do not have the same Y coordinate—that is, when one foot is positioned in front of the other—the distribution is adjusted according to the longitudinal separation between the feet. As this separation increases, a larger portion of the moment about the Z axis is balanced by the X components, while the portion balanced by the pair of Y components decreases.

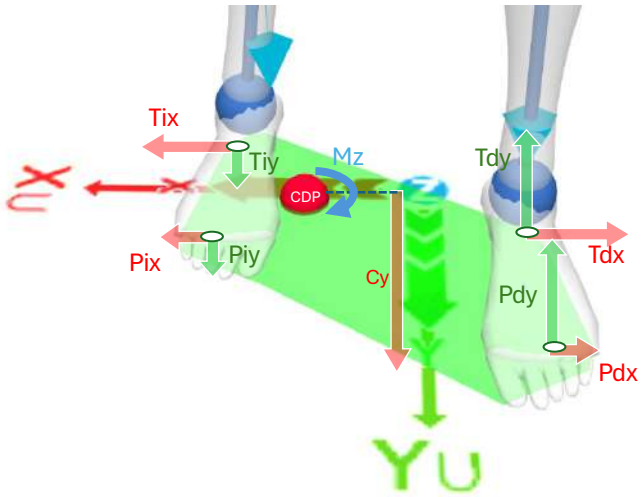


Figure 21: Distribution of horizontal reactions for Y components of external loads.

The case of an external force Cy with a Y component is shown in Figure 21. In this situation, the sum of the Y components of the reactions is equal and opposite to Cy. The X components have a zero resultant and form opposite reactions between both feet:

$$T_{i,x} + P_{i,x} = -(T_{d,x} + P_{d,x})$$

Within each foot, the components are distributed between the heel and the front so that the moment about the ankle is zero. The magnitude of the couple formed by the X components depends on the separation of the feet along the Y axis. When both ankles have the same Y coordinate, these components are zero; their magnitude increases as the longitudinal separation between the feet increases.

7.2 Reaction forces at the foot supports during single-leg stance

Ergoniza Intelligent models two possible point contact areas between each foot and the floor. When the model is in single-leg stance and only one contact area exists between the foot and the floor, all reactions are concentrated at that single support.

When two contact areas exist, the vertical reactions at the heel and forefoot balance the vertical components of body weight and the supported external loads. Their distribution between the supports depends on the position of the center of pressure, as illustrated in Figure 22. The distribution is inversely proportional to the distances from each support to point O, where O is the intersection between the line joining the supports and a line perpendicular to it that passes through the CoP.

The reaction forces at the contacts have horizontal components when the supported external loads also have them. The horizontal reaction forces are calculated separately for the X and Y components.

When two contact areas exist, the distribution of the horizontal components must satisfy both force and moment equilibrium. The sum of the horizontal reactions must be equal and opposite to the horizontal resultant of the external forces. In addition, the moment produced by the loads about the center of the foot must be balanced by the reactions to prevent the foot from rotating.

These conditions alone do not determine a unique solution. Ergoniza Intelligent separates the calculation into the X and Y axes and distributes the moment about the vertical axis between both directions according to the orientation of the foot. When the longitudinal axis of the foot coincides with the model's Y axis, the moment is balanced by the X components of the heel and forefoot reactions; when it coincides with the X axis, the moment is balanced by the Y components. For intermediate orientations, both contributions are combined using continuous weighting based on the square of the cosine of the angle between the foot and the Y axis.

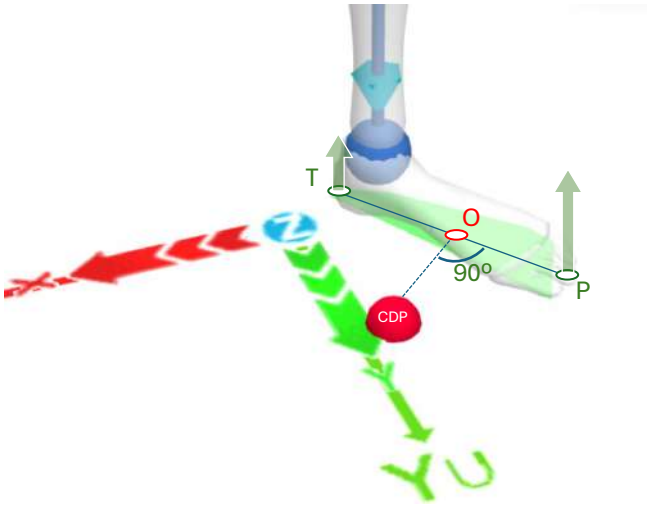


Figure 22: Vertical components of the reaction forces at the supports during single-leg stance.

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7.3 Partial slipping

As shown in Section 6.1, when sustained loads or forces include horizontal components, maintaining the posture requires them to be balanced by the friction forces between the footwear—or another type of support—and the floor. The available friction force must be equal to or greater than the resultant of the horizontal components of the external loads or forces. Otherwise, the model will slip and the posture cannot be maintained.

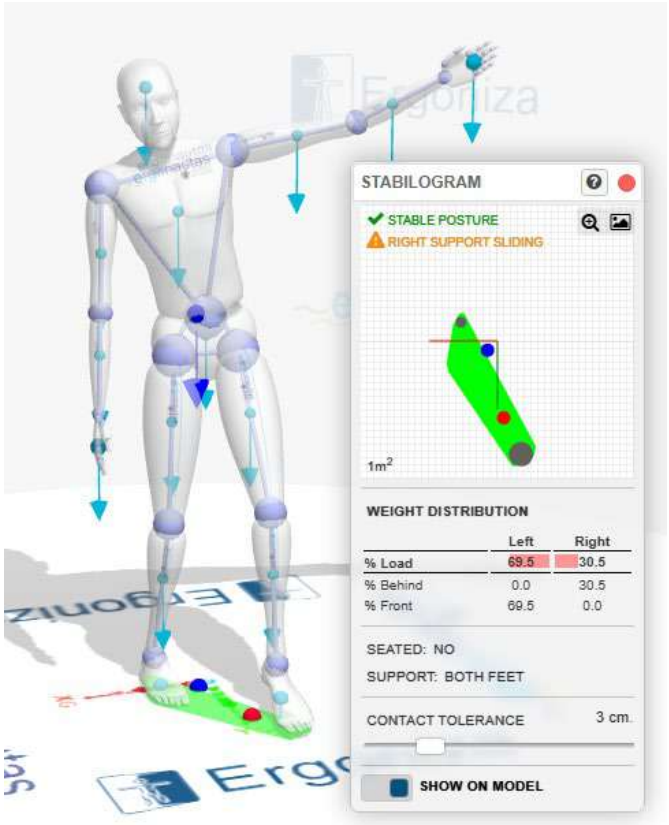


Figure 23: Partial slipping.

After the distribution of the reaction forces in bipedal stance has been calculated, the horizontal resultant assigned to one foot may exceed the maximum friction force available at that support, even though the combined friction capacity of both supports is sufficient to maintain the posture.

Ergoniza Intelligent checks for this situation by calculating the maximum friction force available at each support from the vertical reaction components and the coefficient of friction with the floor, using Equation 1 presented in Section 6.1.

When one support exceeds its friction limit, Ergoniza Intelligent transfers to the other support the portion of the horizontal reaction that it can assume without exceeding its own friction capacity. This correction changes the distribution of the horizontal components between the feet, but not the previously calculated vertical distribution. If the transfer satisfies the global criterion, the **Stabilogram** (Figure 23) displays a partial-slipping warning that identifies the support initially affected, although the posture can be maintained after redistribution.

If the slipping problem at one support cannot be compensated for by redistributing the loads, a global-slipping warning is displayed, indicating that the posture cannot be maintained.

7.4 Reaction moments at the foot supports

Reaction moments may arise when the concentrated forces calculated at the supports are not sufficient by themselves to satisfy moment equilibrium. Ergoniza Intelligent can calculate them during both bipedal and single-leg stance. The simplest case to interpret occurs during single-leg stance when the center of pressure does not lie on the line joining the heel and the front of the foot.

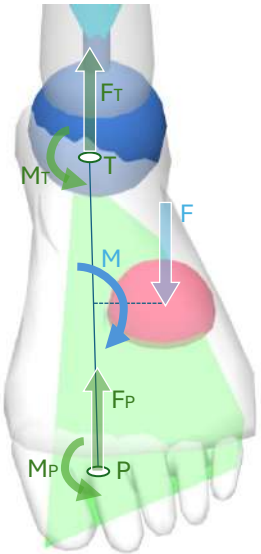


Figure 24: Reaction moments at the supports.

Figure 24 shows a situation of this type in which the model is in a stable posture. Body weight and the vertical components of the external forces and loads are modeled by the resultant force F, applied at the center of pressure. This force is balanced by reactions F_p and F_t at the front of the foot and the heel. Force F also generates a moment M about the axis joining both supports, which must be balanced by reaction moments M_p and M_t. Ergoniza Intelligent distributes M between both supports in proportion to the corresponding vertical reaction force, with $M_p + M_t = M$.

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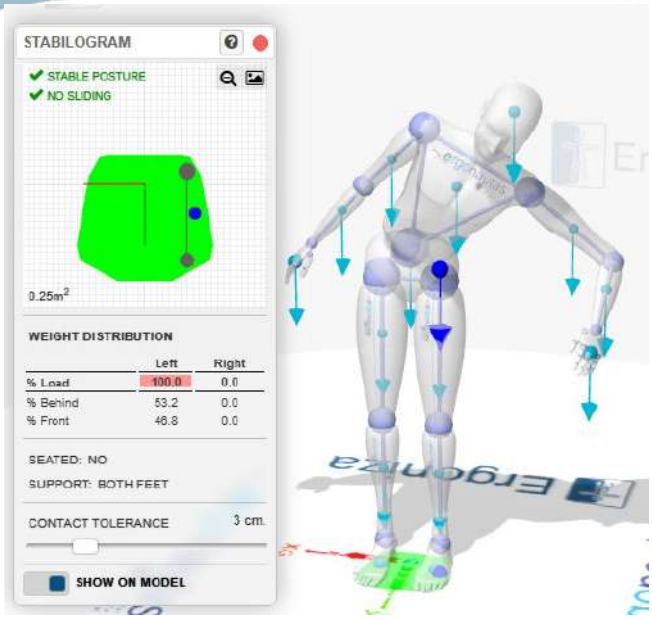


Figure 25: Single-leg stance with both feet in contact with the floor.

It is important to remember that Ergoniza Intelligent considers the model to be in single-leg stance when only one foot is in contact with the floor or when, despite both feet being in contact, the entire load is supported by one of them.

The latter situation is illustrated in Figure 25. The Stabilogram shows that the center of pressure—represented by the blue circle—is not located between the feet, but to the left of the left foot, so the entire load is supported by that foot. Although the right foot remains in contact with the floor, its vertical reaction is zero and it does not effectively participate in supporting the body; the program therefore considers this a single-leg stance.

7.5 Knee supports

Ergoniza Intelligent detects whether one or both knees are supported on the floor. When this occurs, the method used to calculate the reactions at the supports is modified to account for the new contact points.

Figure 26 shows a posture in which the model is supported on both knees and both feet. In this situation, reactions occur at the knees and at one contact area on each foot. Their distribution is calculated in a manner similar to that described above, except that the knees act as the front supports and the foot contact areas as the rear supports, as shown in Figure 27. However, the presence of knee supports changes the calculation of the joint loads at the ankles and knees, as explained in Section 8.2.2.3.

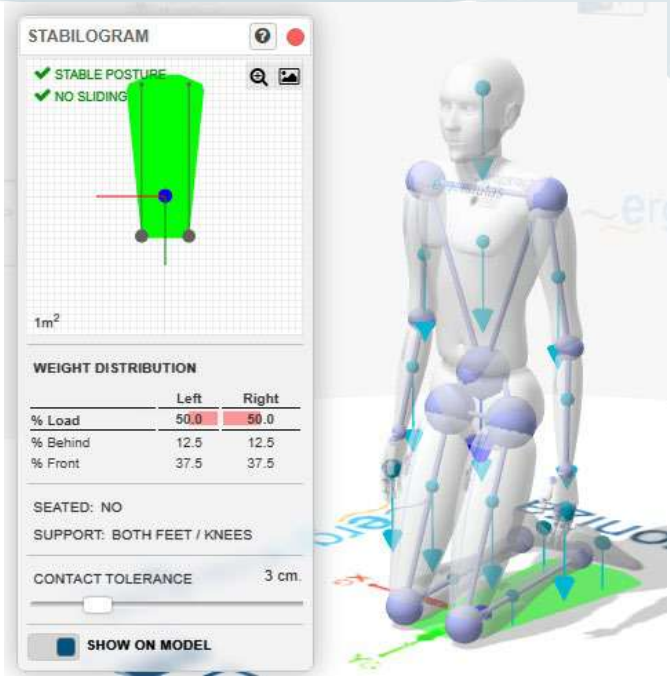


Figure 26: Model supported on the knees and both feet.

Several considerations must be taken into account when knee support occurs. The coefficient of friction used to calculate possible slipping at these supports is the same as that defined for the contact between the sole of the footwear and the floor. In addition, only two support areas can exist on each side of the body. Therefore, if the knee and foot of the same leg are in contact with the floor, the knee and only one area of the foot are considered. When both the forefoot and heel are supported, the program retains the forefoot as the support area; if only the heel is supported, it uses the heel. This simplification generally does not represent a significant limitation, because a posture involving simultaneous contact of the knee, forefoot, and heel on the same side would be extremely forced.

When a knee is supported on the floor, the reaction force exerted by the floor at the contact point must not be confused with the internal reaction at the knee joint. In the situation shown in Figure 27, C_r and C_p identify the knee and foot contacts, while R_{CR} and R_{CP} represent the corresponding reaction forces at those supports. R_R, in turn, represents the reaction transmitted through the knee joint. Although the knee-floor contact point and the joint center are very close, they correspond to different mechanical elements and their forces can differ considerably.

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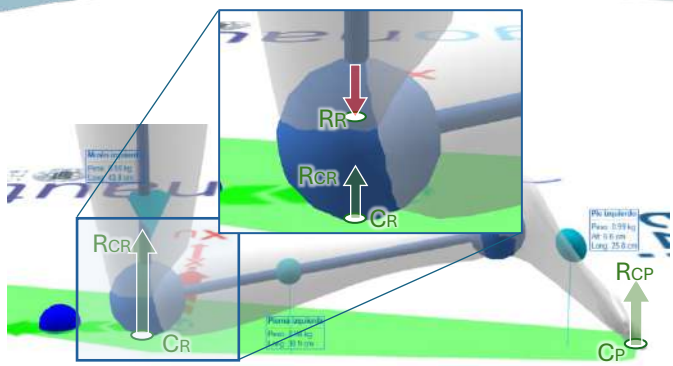


Figure 27: Reactions at the supports when knee contact exists

Accounting for knee supports can give rise to other situations, such as both feet and only one knee being supported, as shown in Figure 28, or both knees and only one foot being supported.

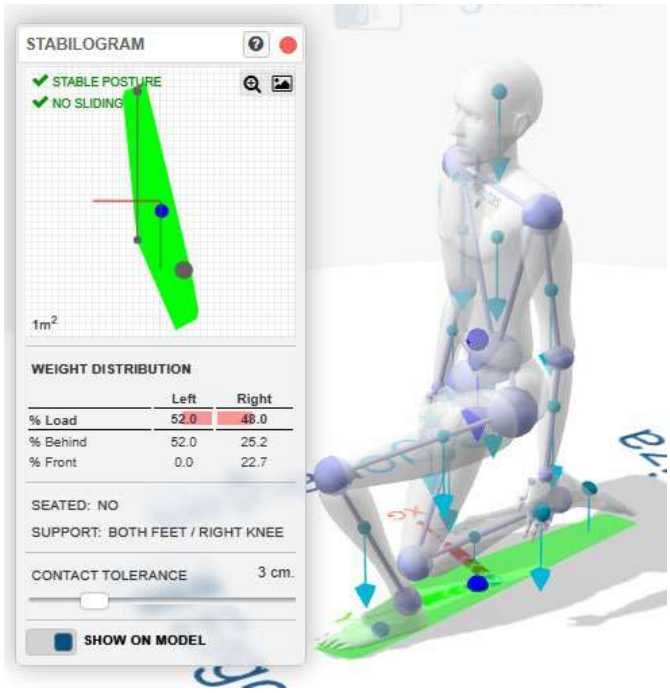


Figure 28: Model supported on one knee and both feet.

7.6 Hip supports

Ergoniza Intelligent detects whether the pelvis is supported on the floor in the area of the ischial tuberosities, as shown in Figure 29. When this occurs, the method used to calculate the reactions at the supports is modified in a manner similar to that described for the knees in Section 7.5. In this case, the rear supports are located at the pelvic contact areas—referred to as hip supports—and the front supports at one contact area on each foot.



Figure 29: Model supported on the hips and both feet.

The limitations described for knee supports also apply to hip supports. The coefficient of friction used to calculate possible slipping is the same as that defined for the contact between the sole of the footwear and the floor. In addition, only two support areas can exist on each side of the body. Therefore, if the hip and foot of the same leg are in contact with the floor, the hip and only one area of the foot are considered. When both the heel and forefoot are supported, the program retains the heel as the support area; if only the forefoot is supported, it uses the forefoot. This simplification generally does not represent a significant limitation, because a posture involving simultaneous contact of the hip, heel, and forefoot on the same side would be extremely forced and uncommon.

As with the knees, the reaction force exerted by the floor on the pelvis must not be confused with the internal reaction at the hip joint. The calculation of joint loads when hip support is present is explained in Section 8.2.2.5.

7.7 Modeling seated posture in a chair

Ergoniza Intelligent distinguishes between a posture seated on the floor and a posture seated on a chair. This distinction is necessary because, although the pelvis may be positioned low relative to the floor in both cases, the mechanical nature of the supports differs. Consequently, the method used to calculate the reactions at the supports and the loads on the lower-limb joints also differs.

When the model is seated on the floor, the reactions at the supports are calculated as described in Sections 7.5 and 7.6 for knee and hip contact with the floor. However, when the posture corresponds to sitting in a chair, Ergoniza Intelligent considers part of the body to be supported by an external surface other than the floor: the chair seat. Under these conditions, although the feet may remain in contact with the

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floor, the distribution of the reactions at the supports and the transmission of loads through the lower kinematic chain differ from those in bipedal stance or a kneeling posture.

7.7.1 Detecting seated posture in a chair

Ergoniza Intelligent considers the model to be seated in a chair under two circumstances. The first is a condition explicitly defined by the user by activating seated posture with the **Seated** button on the left side of the interface. The second uses **automatic detection** based on the posture of the body segments, allowing the program to identify configurations consistent with conventional sitting.

Automatic seated-posture detection can be enabled or disabled under **Model settings** on the right side of the interface. An explicit definition made with the **Seated** button takes precedence over automatic detection when both are enabled.

When the **Seated** button is not active and automatic detection is enabled, Ergoniza Intelligent analyzes the configuration of the pelvis, torso, and lower limbs. A pelvis and torso position consistent with sitting, together with flexion of both hips close to a right angle, allows a conventional seated posture to be identified automatically.

As part of this automatic procedure, if the program detects bilateral hip contact with the floor, it classifies the posture as seated on the floor rather than seated in a chair. Thus, two postures with similar angular configurations but different support conditions can be treated using different mechanical models. If automatic detection is disabled and the **Seated** button is not active, the program will not automatically classify the posture as seated.

7.7.2 Contacts considered when seated in a chair

When the model is seated in a chair, Ergoniza Intelligent establishes four equivalent support points for biomechanical calculation: one at each knee and one at each hip. The points located at the knees do not necessarily represent physical contact between the knees and a surface; rather, they are calculation points used to model load transmission toward the front of each side of the body.

The model therefore assumes two support areas on each side: a front support located at the knee and a rear support located at the hip, in the area of the ischial tuberosity. This simplification allows the same general framework used for other support situations to be applied, representing the contacts as points and linearly distributing the vertical reaction components between them.

The feet may remain in contact with the floor, and this condition continues to be detected by the system. However, when modeling seated posture in a chair, foot contact with the floor does not change the definition of the main supports on each side, which are the corresponding knee and hip. Foot contact is subsequently taken into account when calculating joint loads at the ankles, knees, and hips.

7.7.3 Distribution of reactions at the supports when seated in a chair

When the model is seated in a chair, Ergoniza Intelligent calculates the distribution of the reactions at the supports using the same general principle described in Section 7.1: the distribution of the vertical reaction components depends on the position of the **center of pressure** relative to the **base of support**. The difference is that, in this case, the base of support is constructed from the front and rear supports on each side defined by the knees and hips.

Thus, the distribution of the load between the left and right sides depends on the position of the center of pressure relative to the supports on both sides of the body, while the front/rear distribution within each side depends on the position of the center of pressure between the corresponding knee and hip. Ergoniza Intelligent therefore determines the percentage of the load supported by the left and right sides and, within each side, the percentage of the reaction supported by the front support (knee) and rear support (hip).

This information is represented in the Stabilogram in a manner analogous to that described in Section 7.8. In this case, the front of the support corresponds to the knee and the rear to the hip. The footprint displayed by the system therefore represents the area defined by the supports at both knees and both hips, rather than the plantar contact area used in bipedal stance.

7.7.4 Simplifications assumed when modeling seated posture

Seated posture in a chair introduces a more complex biomechanical situation than bipedal stance or direct support on the floor. In reality, contact between the body and the chair does not occur at points, but over relatively large surfaces: primarily the ischial tuberosities, a variable portion of the back of the thighs, and the backrest, where applicable. In addition, pressure distribution depends on the type of seat, its deformability, trunk posture, and foot position.

The following simplifications are introduced to make the problem manageable within the biomechanical model of Ergoniza Intelligent:

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Equivalent point supports.

The actual pressure distribution on the seat is replaced by point supports located at the knees and hips. This assumption maintains a calculation scheme consistent with that used in the rest of the model.

The actual pressure distribution between the body and the seat is replaced by the equivalent point supports defined in Section 7.7.2: a front support at the knee and a rear support at the hip on each side. This assumption maintains a calculation scheme consistent with that used in the rest of the model.

Two supports on each side of the body.

As in the other situations considered by the system, each side of the body is modeled using a maximum of two support areas. When seated in a chair, these areas correspond to the knee and hip on the respective side.

Partial unloading of the thigh.

When the model is seated in a chair and the foot of one leg is supported, Ergoniza Intelligent considers a substantial portion of the thigh's weight to be supported directly by the seat. In the current formulation of the model, the chair is assumed to support 80% of the thigh's weight, with only the remaining 20% contributing to the reaction force transmitted to the hip. This assumption provides a simplified representation of support under the back of the thigh and avoids overestimating the force transmitted through the hip joint.

Separation between reaction forces and joint moments.

When seated in a chair, Ergoniza Intelligent distinguishes between the reaction force at the hip resulting from the overall equilibrium of the model and the moment that must be supported by the joint itself. This distinction is particularly important in postures with foot support, where the seat relieves some of the joint loading.

When seated in a chair, Ergoniza Intelligent separately calculates the reaction force transmitted through the hip and the moment supported by the joint. When the corresponding foot is supported, the hip may transmit a nonzero force required for model equilibrium, while its joint moment is taken as zero. This assumption is described in greater detail in Section 7.7.5.

7.7.5 Lower-limb joint loads when seated in a chair

In addition to support modeling, sitting in a chair requires specific treatment of the loads at the ankles, knees, and hips. In this case, Ergoniza Intelligent adopts the following calculation assumptions:

Ankle and knee when the foot is supported.

When the model is seated in a chair and the foot of one leg is supported on the floor, Ergoniza Intelligent takes both the joint reaction force and the moment at the ankle and knee of that leg as zero. This simplification does not mean that the external reaction at the plantar support is zero: that reaction is calculated for the overall equilibrium of the model and displayed under *Support reaction loads*. However, in this seated-model condition, it is not transmitted as a joint load through the ankle and knee. The foot support and the support provided by the seat are considered sufficient to unload both joints.

Ankle and knee when the foot is unsupported.

When the model is seated in a chair and the foot of one leg is not supported on the floor, that limb is treated biomechanically as a leg suspended in the air. In this situation, ankle force and moment are calculated from the weight of the foot and the loads applied to it. The knee calculation also includes the weight of and loads on the lower leg. The calculation follows the procedure used for unsupported lower limbs, but without transmitting loads from the upper body to that limb.

Hip when the foot is supported.

When the model is seated in a chair and the foot of one leg is supported, Ergoniza Intelligent retains at the hip the reaction force required for system equilibrium, but takes the joint moment at that hip as zero. This simplification is based on the assumption that the simultaneous presence of the seat and plantar support relieves the hip joint of the need to generate a significant moment to maintain the posture. In this situation, therefore, the hip transmits force but no moment.

Hip when the foot is unsupported.

When the foot of one leg is unsupported, the corresponding hip is calculated using the unsupported-lower-limb branch of the model, accounting for the contribution of the lower limb itself and, where applicable, the loads applied to it, but excluding the direct influence of the upper body. This avoids assigning to that hip loads that are supported by the seat in a seated posture.

7.7.6 Consideration of the chair backrest

Ergoniza Intelligent can also account for the presence of a chair backrest. Posterior trunk support changes the transmission of loads between the body and the chair. Experimental studies using adjustable lumbar supports have observed changes in

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load distribution, reduced lumbar muscle activity, and better maintenance of lumbar lordosis while seated[8].

Ergoniza Intelligent allows the user to indicate that the trunk is supported by a backrest using the **Backrest** control under **Model settings** on the right side of the interface. This option should be enabled only when the model is seated in a chair and the trunk is actually supported by the backrest.

When the backrest is enabled and the posture has been identified as seated in a chair, Ergoniza Intelligent analyzes the local components of the lumbar moment separately. If the flexion-extension component corresponds to a flexion demand that can be relieved by posterior support, the system considers this component to be supported by the backrest and sets it to zero. The lateral-flexion and rotation components are not changed.

The same criterion is applied independently to the flexion-extension component of the moment calculated at the L4/L5 joint. From the corrected L4/L5 moment, Ergoniza Intelligent calculates muscle activity, compression force, and shear forces in the intervertebral disc. Compression is not set to zero because the weights of the body segments, external loads, and the muscle forces required to balance the lateral-flexion and rotation components continue to act. The general procedure used to calculate these results is described in Section 12.

When the flexion-extension component corresponds to extension—a situation in which the trunk does not tend to rest against the backrest—the system applies no unloading. It also does not change the results if the backrest is disabled or if the posture is seated on the floor.

This approach provides a simplified representation of the unloading effect of a backrest in seated postures with posterior trunk support, avoiding the attribution to the trunk muscles of demands that are actually supported by the chair structure. The model does not explicitly calculate the force exerted by the backrest or its point of application. Therefore, enabling the backrest does not alter the reaction forces or their distribution among the other supports. Its effect is represented exclusively by unloading the local flexion component of the lumbar moments.

7.7.7 Scope and limitations of the seated-posture model

The modeling of seated posture in a chair implemented in Ergoniza Intelligent should be understood as a simplified biomechanical approximation. Its purpose is to provide an operational representation consistent with the rest of the system while retaining the ability to analyze stability, the distribution of reactions at the supports, and joint loads.

However, this model does not reproduce all the actual mechanical phenomena associated with interaction between the body and a chair. In particular:

- the pressure distribution over the seat or backrest is not represented continuously;
- deformation of the seat and soft tissues is not modeled explicitly;
- geometric or mechanical differences among chair types are not considered;
- partial unloading of the thigh on the seat is represented using a fixed, simplified assumption;
- the reaction force exerted by the backrest and its point of application are not calculated; its effect is represented exclusively by unloading the flexion component of the lumbar moments.

Therefore, the results should be interpreted as a reasonable biomechanical estimate of seated posture in a chair, useful for comparative ergonomic assessment and trend analysis, but not as an exact representation of the actual mechanical interaction between the body and the furniture.

7.8 Stabilogram and distribution of reactions at the supports

Ergoniza Intelligent presents the results of the support-reaction calculation under **Support reaction loads** in the **Exertion** window and in the **Stabilogram** tool (Figure 30). The former shows the distribution of forces, expressed in newtons (N), and reaction moments, expressed in newton-meters (N·m), between the left and right supports and between the front and rear of each support.

The interpretation of the front and rear areas depends on the support type. When support occurs at the feet, the front corresponds to the forefoot and the rear to the heel. When one leg is supported simultaneously at the knee and foot, the front corresponds to the knee and the rear to the forefoot. When

[8] Mohsen Makhsous et al., "Sitting with Adjustable Ischial and Back Supports: Biomechanical Changes," *Spine* 28, no. 11 (2003): 1121–22, doi:10.1097/01.BRS.0000068243.63203.A8.

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modeling seated posture in a chair, the equivalent front support corresponds to the knee and the rear support to the hip on the same side.

In the example in Figure 30, the supports correspond to the feet and the reaction forces have only a vertical component because the model is not supporting loads with horizontal components. In this particular case, the distribution of forces relative to the center of pressure satisfies equilibrium without reaction moments arising at the supports.

The bottom of the **Support reaction loads** section also displays the **Residual moments**, which are zero in the example in Figure 30. Residual moments are discussed in Section 8.3.

The Stabilogram shows the position of the center of pressure, represented by a blue circle, relative to the base of support. It also represents the support points and the geometry defining the base. The size of the markers associated with the support points is proportional to the reaction supported by each point.

Below the graph, the percentage distribution of the load between the left and right supports is shown, together with its internal distribution between the front and rear areas. The interpretation of these areas depends on the support type, as indicated above.

Ergoniza Intelligent estimates physical contact of the feet, knees, and hips with the floor from their distance to it. The **Contact tolerance** control sets the maximum distance at which contact is considered to exist. When seated in a chair, the equivalent knee and hip supports are established according to the model described in Section 7.7, while foot contact with the floor continues to be determined from foot position.

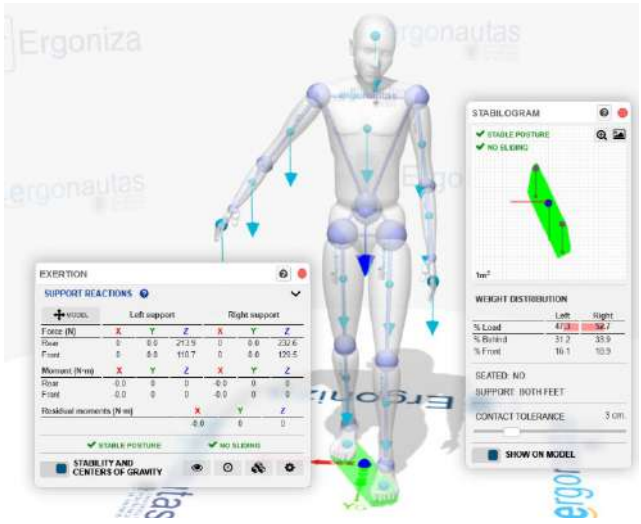


Figure 30: Stabilogram and distribution of reaction loads at the supports.

8 JOINT LOAD CALCULATION

When maintaining a posture, the joints must support loads caused by the weight of the body segments and by external forces, loads, or moments applied to the model. These loads are transmitted along the kinematic chains and appear at the joints as forces and moments.

For equilibrium to exist—assuming that the posture remains static—reaction forces and moments arise at the joints with the same magnitude as, and in the direction opposite to, the loads they support.

For equilibrium to exist, assuming that the posture remains static, the sum of the forces and moments must be zero. Therefore, a reaction force and moment arise at each joint with the same magnitude as, and in the direction opposite to, the resultant of the loads transmitted by the body segments.

The moment generated by a force about a joint is obtained from the cross product of the position vector from the joint to the point of application of the force and the force vector. Using this relationship and the equilibrium equations, Ergoniza Intelligent calculates the forces and moments transmitted by each joint.

Figure 31 uses the wrist and elbow joints as examples. The external load (C) and the hand's own weight (Pp_m) act on the hand. To simplify the representation, both forces are considered to act at the hand's center of gravity (CdG_m).

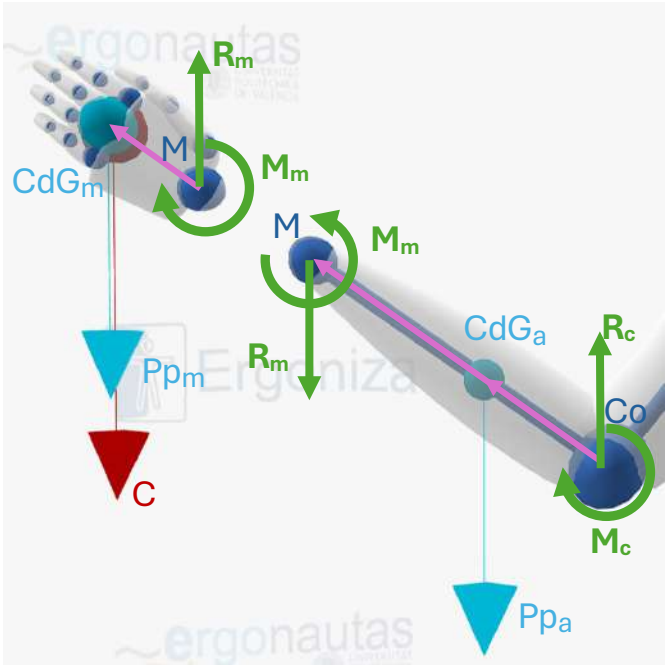


Figure 31: Calculation of the forces and moments transmitted by the wrist and elbow joints.

In the following equations, $\vec{r}_{A\to B}$ represents the position vector from point A to point B. The calculation uses the resultant force

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and moment transmitted from the distal segment to the proximal segment; the reactions that maintain equilibrium have the same magnitude and opposite direction. At the wrist, the resultant force Rm is the sum of the external load C and the hand's own weight Pp_m, while M_m is obtained by adding the moments produced by both forces about the joint. These relationships are expressed in Equation 2.

$$\overrightarrow{Rm} = \vec{C} + \overrightarrow{Pp_m}$$

$$\overrightarrow{Mm} = \vec{r}_{M\rightarrow CdG_m} \times \vec{C} + \vec{r}_{M\rightarrow CdG_m} \times \overrightarrow{Pp_m}$$

Equation 2

A similar procedure is used to calculate the force R_c and moment M_c transmitted by the elbow, now analyzing the forearm. This segment is acted on by the load transmitted from the hand through the wrist, the forearm's own weight (Pp_a), and the moment transmitted by the wrist (M_m).

Again assuming that the posture remains static and applying the laws of equilibrium, the values of R_c and M_c can be determined using Equation 3.

When the equilibrium equations are applied to the forearm, the resultant force R_c is obtained by adding the load transmitted by the wrist and the forearm's own weight. Moment M_c also includes the moment transmitted by the wrist and the moments produced by both forces about the elbow. These relationships are expressed in Equation 3.

$$\overrightarrow{Rc} = \overrightarrow{Rm} + \overrightarrow{Pp_a}$$

$$\overrightarrow{Mc} = \overrightarrow{Mm} + \vec{r}_{Co\rightarrow M} \times \overrightarrow{Rm} + \vec{r}_{Co\rightarrow CdG_a} \times \overrightarrow{Pp_a}$$

Equation 3

This procedure can be applied successively along the entire kinematic chain. Once the force and moment transmitted by the elbow are known, those corresponding to the shoulder can be calculated, and the same process can continue with the joints located closer to the torso.

8.1 Upper-limb joint loads

Ergoniza Intelligent begins calculating joint loads at the wrists. It then applies the procedure described in the preceding section to calculate the loads transmitted through the elbows and shoulders. At each step, the weight of the corresponding segment and any external loads acting on it are included. Finally, the calculation of loads at the L5/S1 joint combines the forces and moments transmitted through both shoulders with the weight of the torso and head, as well as external loads applied to the torso, head, neck, or upper limbs.

8.2 Lower-limb joint loads

The calculation of lower-limb joint loads at the hips, knees, and ankles depends on the supports detected. Ergoniza Intelligent distinguishes among an unsupported model, unilateral support, bilateral support, and sitting in a chair. It also adapts the calculation sequence according to whether contact occurs through the feet, knees, hips, or seat.

8.2.1 Lower-limb joint loads in an unsupported model

When there is no contact with the floor, the model is unbalanced because no reaction forces at the supports counteract the effect of gravity.

A static analysis of the whole body cannot be performed under these conditions. However, Ergoniza Intelligent calculates and displays the joint loads by assuming that the pelvis remains fixed in space. The results are equivalent to those for a situation in which the model is held static by a rigid restraint applied at the pelvis, a circumstance that must be taken into account when interpreting them.

When the model remains unsupported, loads at the wrists, elbows, shoulders, and L5/S1 joint are calculated as described in Section 8.1. Lower-limb joint loads are calculated analogously, but starting at the ankles and proceeding upward through the knees to the hips.

This virtual fixation of the pelvis interrupts load transmission between the torso and the lower limbs. Consequently, the forces and moments calculated at the L5/S1 joint are not transmitted to the hips, and the loads on each leg are calculated independently.

8.2.2 Lower-limb joint loads in a supported model

When supports are present, the calculation procedure depends on the number and type of contacts detected. The system first distinguishes between unilateral and bilateral support. It then adapts the calculation sequence according to whether contact occurs through the feet, knees, or hips. Sitting in a chair is handled using a specific procedure.

With unilateral support—for example, during single-leg stance—one side of the body transmits no reaction to the floor. Consequently, loads from the upper body and the unsupported lower limb must be transmitted through the supported side to the contact point.

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Loads from the upper body and unsupported lower limb converge at the pelvis. From there, the calculation proceeds downward through the supported limb, as shown in Figure 32 (A). When support occurs at the foot, the sequence passes through the hip, knee, and ankle. If support occurs at the knee or hip, the chain ends at that contact area. Finally, the loads at the joint connected to the support are used to determine the corresponding reactions.

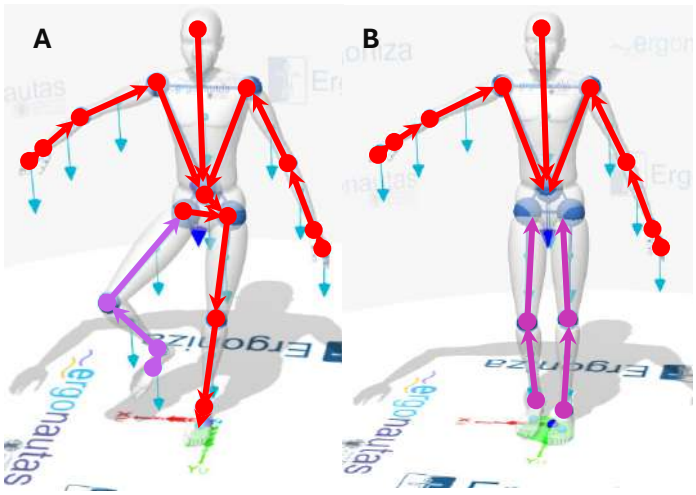


Figure 32: Sequence for calculating joint loads during single-leg stance (A) and bipedal stance (B).

When the model is in bipedal stance, both legs are in contact with the floor and loads from the upper body must be distributed between them. Under these conditions, the equilibrium equations alone are insufficient to determine how all reactions are distributed between the two lower limbs. The system is therefore statically indeterminate and requires an additional procedure to calculate the joint loads.

During bilateral support, Ergoniza Intelligent first determines the reaction forces and moments at the supports using the procedure described in Section 7. It uses the position of the center of pressure relative to the base of support and the distribution of the external loads. When only the weights of the body segments act, the position of the resultant is obtained from the body center of mass. Once the reactions at the supports are known, the program calculates the loads at the ankles, knees, and finally the hips. The calculation therefore proceeds from the supports toward the pelvis, as shown in Figure 32 (B).

The specific calculation method depends on the support type. The following sections describe the procedures applied when contact occurs through the feet, knees, or hips, as well as the particular considerations associated with postures involving extreme flexion.

When the model is seated in a chair, the seat acts as an additional support for the pelvis. Ergoniza Intelligent then applies a specific procedure that accounts for whether each foot is in contact with the floor and adapts load transmission through the lower-limb joints. The model used to represent support from the seat and backrest is described in Section 7.7.6.

8.2.2.1 Calculating ankle loads with foot support

When support on the analyzed side occurs only at the foot, with no simultaneous knee or hip contact with the floor, the known reactions at the foot supports allow the ankle load to be calculated. Following the notation shown in Figure 33, this load consists of force F_{Tob}, with components F_{Tobx}, F_{Toby}, and F_{Tobz}, and moment M_{Tob} (M_{Tobx}, M_{Toby}, and M_{Tobz}). Because this is a static system, the sum of the forces and moments acting on the foot must be zero.

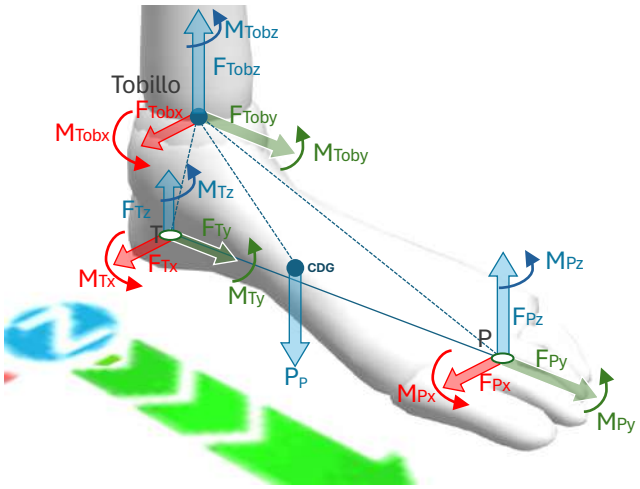


Figure 33: Calculation of ankle loads.

To establish moment equilibrium about the ankle, the moments generated by the reaction at the heel (F_T), the reaction at the forefoot (F_P), and the foot's own weight (P_P) are considered. Where present, the reaction moments calculated at the heel and forefoot supports, represented by M_T and M_P, respectively, are also included. These equilibrium conditions are expressed in Equation 4.

$$\overrightarrow{F_{Tob}} + \overrightarrow{F_T} + \overrightarrow{F_P} + \overrightarrow{P_P} = \vec{0}$$

$$\overrightarrow{M_{Tob}} + \vec{r}_{Tob\to T} \times \overrightarrow{F_T} + \vec{r}_{Tob\to P} \times \overrightarrow{F_P} + \vec{r}_{Tob\to CdG} \times \overrightarrow{P_P} + \\ + \overrightarrow{M_T} + \overrightarrow{M_P} = \vec{0}$$

Equation 4

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8.2.2.2 Calculating knee and hip loads with foot support

When support occurs exclusively at the feet, once the forces and moments transmitted through the ankles are known, the loads at the knees can be calculated using the same procedure presented at the beginning of Section 8. This calculation includes the load transmitted by the ankle, the lower leg's own weight, and any external loads applied to this segment. The loads calculated at the knees, together with the weight of the thighs and the corresponding external loads, are then used to obtain the loads at the hips. The calculation therefore proceeds from the supports toward the pelvis by successively applying the equilibrium equations.

8.2.2.3 Calculating ankle and knee loads when knee support is present

As explained in Section 7.5, Ergoniza Intelligent detects knee contact with the floor and calculates the corresponding reaction forces. When this type of contact exists, the joint loads no longer depend only on the reactions at the foot supports, but also on the reaction exerted by the floor on the knee. Therefore, the calculation of ankle and knee loads differs from the procedure described in Sections 8.2.2.1 and 8.2.2.2.

As an example, Figure 34 shows a leg supported simultaneously at the knee and forefoot. The same procedure can be applied to the opposite side and to other equivalent configurations. To facilitate interpretation, the figure shows only the reaction forces at the supports and the joint loads; moments are not included.

Two reaction forces appear in the situation shown: R_R, applied at the knee support, and R_P, applied at the forefoot. The load F_T calculated at the ankle must balance reaction R_P and the foot's own weight (PP_{PIE}). The ankle moment is obtained by considering the moments produced by these forces about the joint and, where applicable, the reaction moments calculated at the supports.

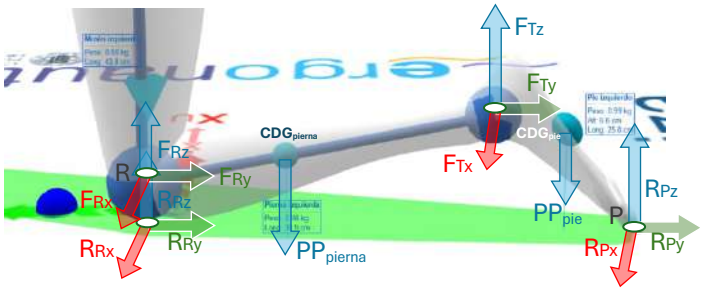


Figure 34: Forces considered when calculating ankle and knee loads in the presence of knee support.

The force F_R calculated at the knee must balance the load transmitted by the ankle, the lower leg's own weight (PP_{Pierna}), and the reaction (R_R) exerted by the floor on the knee. In the simplification used by Ergoniza Intelligent, R_R is considered to be applied at the joint center and therefore generates no moment about it. The knee moment is obtained from the moment transmitted by the ankle and the moments produced by the ankle load and the lower leg's own weight.

8.2.2.4 Treatment of the knee joint moment under extreme lower-leg flexion

When the feet and/or knees are in contact with the floor and flexion of the lower legs relative to the thighs is extreme, the thighs are supported on the lower legs. Because of this support, the muscles acting at the knee joint do not need to overcome the moment associated with loads located higher in the kinematic chain.

When the foot is supported on the floor and knee flexion is very pronounced, Ergoniza Intelligent assumes that the back of the thigh contacts the lower leg and provides passive support. Under this simplification, the moment generated by loads located above the knee is not attributed to the muscles acting on the joint. Contact between the thigh and lower leg is inferred from the flexion angle and the presence of foot support.

Figure 35 shows two configurations consistent with this simplification. In the main image, the model adopts a squatting posture in which the backs of the thighs can rest on the lower legs. The detail image shows that this situation can also coexist with knee support on the floor, provided that the corresponding foot remains in contact.

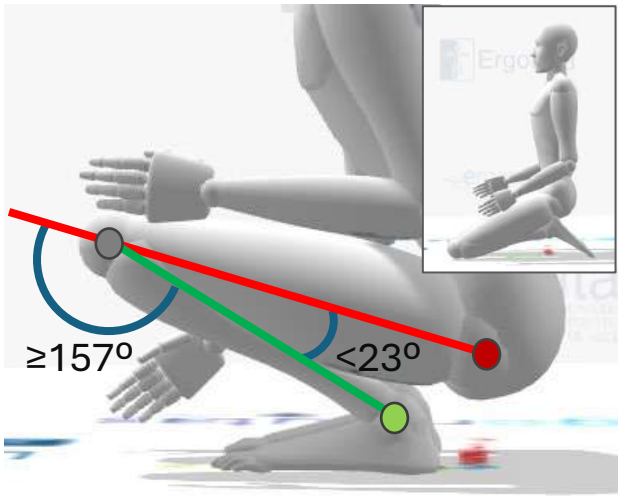


Figure 35: Extreme-flexion situations in which the model sets the knee joint moment to zero.

When the foot is in contact with the floor and the knee flexion angle is greater than 157°, Ergoniza Intelligent sets the

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calculated knee joint moment to zero. This condition can coexist with support of the knee itself on the floor. The joint force is not set to zero and continues to be calculated using the equilibrium equations.

Even though Ergoniza Intelligent sets the load moment to zero in this situation, the posture should be analyzed carefully. Setting the moment to zero is a model simplification and does not mean that the knee is unloaded. The joint force continues to act, and neither the contact pressures nor the local distribution of loads over the cartilage, menisci, or ligaments are calculated explicitly.

Epidemiological studies have found an association between cumulative exposure to kneeling or squatting tasks and a higher frequency of symptomatic knee osteoarthritis. This association may be stronger when these postures are maintained or repeated for prolonged periods and combined with handling additional loads[9].

Consequently, a zero joint moment in this case must not be interpreted as an absence of load or ergonomic risk. The calculated knee force and the results of postural assessment methods should be considered together when analyzing this type of posture.

8.2.2.5 Calculating lower-limb joint loads when hip support is present

When the hip on the analyzed side contacts the floor, a reaction force arises at that point and changes the calculation of the corresponding lower-limb joint loads. The procedure differs from that described for support exclusively at the feet in Sections 8.2.2.1 and 8.2.2.2, and is analogous to that used when knee support is present, as described in Section 8.2.2.3.

As an example, Figure 36 shows a leg supported simultaneously at the heel and hip. The same procedure can be applied to the opposite side and to other equivalent configurations. To facilitate interpretation, the figure shows the reaction forces at the supports and the joint loads; moments are not included.

Two reaction forces appear in the situation shown: R_T, applied at the heel, and R_C, applied at the hip. Both are represented by their X, Y, and Z components. The joint forces calculated at the ankle, knee, and hip are also shown.

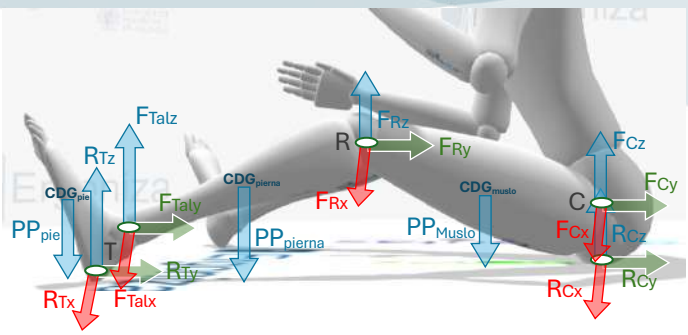


Figure 36: Forces considered when calculating ankle, knee, and hip loads in the presence of hip support.

In this situation, the ankle load F_{Tal} corresponds to the combination of the foot's own weight (PP_{PIE}) and the reaction at the heel R_T, accounting for their opposite signs. The reaction moment at the ankle—not shown in the image—is the moment about the ankle produced by R_T and PP_{PIE}.

The joint force F_{Tal} calculated at the ankle must balance the reaction R_T exerted by the floor on the heel and the foot's own weight, PP_{PIE}. The ankle moment is obtained by considering the moments produced by these forces about the joint. Where applicable, the residual moments described in Section 8.3 are incorporated into the result.

The joint force F_R calculated at the knee must balance the load transmitted by the ankle, the lower leg's own weight, PP_{Pierna}, and any external loads applied to this segment. The knee moment is obtained from the moment transmitted by the ankle and the moments produced by these forces about the joint.

Finally, the joint force F_C calculated at the hip must balance the load transmitted by the knee, the thigh's own weight, PP_{Muslo}, any external loads applied to this segment, and the reaction R_C exerted by the floor on the hip. Ergoniza Intelligent considers R_C to be applied at the joint center, so it generates no moment about the joint. The hip moment is obtained from the moment transmitted by the knee and the moments produced by the remaining forces.

8.3 Residual moments

When support is distributed between both sides of the body, Ergoniza Intelligent calculates loads along the upper chain, beginning at the wrists and proceeding successively through the elbows, shoulders, and L5/S1 joint. This procedure

[9] A. Klussmann et al., "Individual and Occupational Risk Factors for Knee Osteoarthritis: Results of a Case-Control Study in Germany," *Arthritis Research and Therapy* 12, no. 3 (2010): R88, doi:10.1186/ar3015; A. Seidler et al., "The Role of Cumulative Physical Work Load in Symptomatic Knee Osteoarthritis - A Case-Control Study in Germany," *Journal of Occupational Medicine and Toxicology* 7, no. 1 (2008): 21, doi:10.1186/1745-6673-3-14.

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determines the force and moment transmitted to the pelvis from the upper body.

Independently, loads along the lower chain are calculated from the reaction forces determined at the contacts with the floor, following the procedure described in Section 7. The calculation proceeds from the ankles toward the knees and hips, as shown in Figure 32 (B), and determines the actions transmitted to the pelvis from the lower limbs.

Under static-equilibrium conditions, the resultant moment at the pelvis obtained from the upper chain should be compatible with that obtained from the lower limbs. However, the distribution of reactions among the different supports and the simplifications used to represent the contacts may produce a discrepancy between the two results. Ergoniza Intelligent uses this difference to determine the residual moments. For rotation about the vertical axis, the moment produced by the horizontal components of the external forces is also taken into account.

Once calculated, the residual moments are distributed among the active support areas in proportion to the portion of the total load supported by each one. These moments are incorporated into the calculation of the lower-limb joints to close the overall equilibrium of the model. They do not represent newly measured external loads, but rather compensation terms derived from the calculation procedure. Their values are shown under **Residual moments** in the Support reaction loads panel (Figure 30) described in Section 7.8.

8.4 Joint loads

Ergoniza Intelligent displays the X, Y, and Z components of the resultant force and moment acting at each joint under the analysis conditions under **Joint loads** in the **Exertion** window (Figure 37). Forces are expressed in newtons and moments in newton-meters.

EXERTION

JOINT LOADS

MODEL | Moments (N·m) | | | Forces (N) | | |

Joint	X	Y	Z	X	Y	Z
Left wrist	0.9	0.7	0.0	-8.4	12.1	-4.2
Right wrist	0.9	0.7	0.2	-8.4	12.1	-4.2
Left elbow	3.5	2.5	0.0	-8.4	12.1	-16.1
Right elbow	3.5	3.1	1.0	-8.4	12.1	-16.1
Left shoulder	8.1	4.1	-0.7	-8.4	12.1	-35.8
Right shoulder	8.1	6.8	1.2	-8.4	12.1	-35.8
Lumbar L5/S1	7.2	4.3	0.4	-16.9	24.1	-358.4
Left hip	-2.3	8.9	0.5	-10.3	8.5	-291.6
Right hip	7.8	5.7	0.3	-6.6	15.6	-142.6
Left knee	-3.4	4.4	0.4	-10.3	8.5	-378.9
Right knee	2.5	2.8	0.3	-6.6	15.6	-229.9
Left ankle	-14.7	0.4	0.7	-10.3	8.5	-408.1
Right ankle	-8.6	0.3	0.4	-6.6	15.6	-259.2

STABLE POSTURE NO SLIDING

STABILITY AND CENTERS OF GRAVITY

Figure 37: Joint loads displayed in the Exertion window.

By default, forces and moments are expressed relative to the **model reference system** (see Section 3.2). The text on the button at the top left of the table indicates the system currently in use. Pressing **Model** changes the representation to the **global reference system** (see Section 3.1), and the button then displays Global; pressing it again restores the model reference system. When the model is rotated about the vertical axis, the horizontal X and Y components expressed in the global reference system may change, while the Z axis continues to represent the vertical direction.

The sign of each component indicates the direction of the force or moment relative to the axes of the selected reference system. In a limb supported on the floor, the Z component of the joint forces is usually negative because the displayed internal resultant must balance the upward reaction at the support after accounting for the weights of the segments below each joint.

When a limb is unsupported, no floor reaction is transmitted through it. In this situation, the Z component of its joint forces may be positive because the proximal joints must support the weight of the distal segments and any loads applied to them.

The sign indicates only the direction of the resultant under the convention used and must not be interpreted by itself as a measure of muscular exertion or ergonomic risk. Muscular demand is assessed primarily from joint moments and the corresponding percentage exertion indicators.

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8.5 Limitations in calculating moments and biomechanical exertions

Joint moments and the exertions performed are calculated using a static model. This means that the effects of acceleration are not taken into account. Therefore, the biomechanical model results apply primarily to slow movements in which acceleration effects are negligible. In other words, Ergoniza Intelligent analyzes movements as a sequence of static postures, so the results are accurate only when there are no sudden movements or substantial accelerations.

9 ASSESSMENT OF ERGONOMIC RISK FACTORS

Ergoniza Intelligent can assess different ergonomic risk factors associated with posture, physical exertion, and the mechanical interaction between the body and its environment. Among other aspects, the system can estimate the joint exertions required to maintain a posture or support a load, the discomfort associated with unsuitable or sustained postures, postural load using observational methods such as REBA or LUBA, the compression and shear forces acting on the L4/L5 intervertebral disc, and the risk associated with lifting or lowering loads. It can also analyze postural stability, the possibility of slipping, and weight distribution among the different supports.

These results should not be interpreted as alternative measures of the same quantity, but as complementary approaches to different dimensions of ergonomic risk. Each corresponds to a specific mechanism and is calculated using its own models, assumptions, and criteria. For example, joint-exertion analysis estimates the mechanical demand required to maintain a posture or support a load; postural-discomfort analysis assesses the deviation of body segments from functional or comfortable zones; and lumbar-compression analysis estimates the mechanical loads on the L4/L5 region. Stability analysis checks whether the projection considered by the model remains within the polygon defined by the supports, while slipping analysis compares the required horizontal reactions with the available friction capacity. Weight distribution provides additional information about the load supported by each support, but does not by itself constitute a direct measure of risk.

Although these factors are calculated separately, they are not strictly independent. In many situations, relationships exist among them. A posture involving pronounced trunk inclination can simultaneously increase postural discomfort, moments in the lumbar region, and compression force on the intervertebral discs. A heavy load held with the arms away from the body can increase exertion at the shoulders, elbows, and wrists, change the position of the center of pressure, and affect stability. Similarly, a postural change intended to reduce the joint moment in one area of the body may increase the load or postural demand on other segments.

For this reason, Ergoniza Intelligent does not reduce the assessment to a single overall risk indicator. Combining all factors into a single score could conceal relevant information about the source of the ergonomic problem and make it more difficult to identify appropriate corrective measures. Two tasks with the same overall value could have very different risk profiles: one dominated by overexertion, another by extreme postures, another by lumbar compression, and another by instability. The value of the analysis lies precisely in distinguishing these dimensions and allowing the assessor to identify which factors contribute to the observed risk.

Assessing the indicators separately also facilitates technical interpretation of the results. If the main problem is joint exertion, preventive measures can focus on reducing the load, bringing it closer to the body, modifying the grip, or introducing mechanical aids. If the dominant factor is postural discomfort, improvements can focus on redesigning working heights, reaches, work orientations, or rest-break requirements.

If the main factor is lumbar compression, particular attention should be given to the magnitude and position of the loads, trunk flexion and asymmetry, and, where applicable, the conditions under which lifting or lowering is performed. If instability or a highly asymmetric weight distribution is detected, it may be necessary to modify the base of support, posture, or point of application of the forces. When a slipping risk exists, the applied horizontal forces, coefficient of friction, and condition of the footwear and support surface should also be reviewed.

Consequently, the results provided by Ergoniza Intelligent should be interpreted as a structured set of biomechanical and ergonomic results and estimates, not as an automatic decision. The system helps identify, quantify, and compare risk factors, but the final ergonomic assessment remains the responsibility of the professional analyzing the task. The assessor must consider the results together with observation of the actual activity, the duration and frequency of exposure, the characteristics of the worker population, organizational conditions, and any other relevant information about the job.

The absence of high values for a particular indicator does not necessarily imply the absence of overall risk. Similarly, a high value in a specific dimension must be interpreted in its context

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and in relation to the other factors. The role of Ergoniza Intelligent is to provide information that supports professional judgment, facilitates comparison among design alternatives, and guides preventive decision-making.

The following chapters address the assessment of each risk factor independently. Nevertheless, the assessor must understand the relationships among them and consider them together when forming a technical judgment.

10 OVEREXERTION RISK ANALYSIS

The procedure described in Section 8 makes it possible to estimate the joint loads (reaction forces and moments) associated with maintaining a posture, with or without external loads. To assess exertion demands, Ergoniza Intelligent uses the required joint moments and compares them with the estimated moment-generating capacities and the applicable reference limits. Reaction forces are displayed as supplementary mechanical information, but they do not directly enter into the calculation of the exertion indicators described in this section.

10.1 Maximum joint capacities

In a simplified biomechanical analysis, reaction forces represent the resultant of the actions transmitted through the articular surfaces and the structures that stabilize the joint. Joint moments must be balanced primarily by the muscle groups acting on the joint, although passive structures (such as ligaments, joint capsules, and periarticular tissues) may also contribute, particularly when the joint is near the limits of its range of motion. Moment-generating capacity depends on numerous factors, which makes estimating its maximum value a complex task.

The muscles acting on a joint perform different functions and do not have the same spatial arrangement. During elbow flexion, for example, the flexor muscles act as agonists and generate the primary moment, while the extensors act as antagonists. The latter may reduce their activation but do not necessarily relax completely, because coactivation of agonist and antagonist groups helps stabilize the joint.

A muscle's force-generating capacity also varies with its degree of lengthening. In addition, as movement occurs, the position of its line of action changes relative to the center of the joint and therefore so does the moment arm through which it acts. Muscle and tendon attachments are also distributed over a given surface and cannot always be represented as simple geometric points. Finally, muscular capacity can vary considerably even among people with similar body builds.

To assess joint exertion, Ergoniza Intelligent does not model each muscle individually or explicitly calculate coactivation between agonists and antagonists. The program groups muscular actions into functional muscle groups and, based on the direction of the required joint moment, identifies the group that must generate it.

Maximum Voluntary Contraction (MVC) is the maximum exertion that a person can voluntarily produce with a muscle or muscle group under specified conditions. In the biomechanical model of Ergoniza Intelligent, this capacity is represented by the maximum joint moment that the corresponding functional muscle group can generate. This value represents an estimated maximum capacity and does not, by itself, constitute a recommended limit for performing a task.

Essentially, in a static or quasi-static posture, the magnitude of the joint moment required to balance the weight of the body segments and the external loads must be lower than the magnitude of the moment-generating capacity of the active muscle group. When the required moment approaches that maximum capacity, the relative exertion demand is high, even if the action can be maintained only for a very short time.

Ergoniza Intelligent bases its calculation of maximum moment-generating capacities on several studies[10], from which regression equations were developed to estimate the capacity of the different functional muscle groups. In total, Ergoniza Intelligent uses 21 mathematical models: six for the wrist (flexion, extension, radial deviation, ulnar deviation, pronation, and supination); two for the elbow (flexion and extension); six for the shoulder (abduction, adduction, anterior rotation, posterior rotation, medial humeral rotation, and lateral humeral rotation); four for the L5/S1 intervertebral joint (flexion, extension, lateral flexion, and rotation); and three for flexion/extension of the hip, knee, and ankle. For modeling purposes, radioulnar rotation that produces forearm pronation and supination is included in the wrist functional block. Similarly, the ankle model incorporates the corresponding functional movements of the foot.

The variables considered are not identical in all equations. Depending on the joint and movement being analyzed, the models may incorporate the posture of the assessed joint and adjacent joints, sex, age, and laterality or dominance. The

[10] H. H Clarke, *Muscular Strength and Endurance in Man* (Englewood Cliffs, NJ: Prentice Hall, 1966); F. Schanne, "A Three-Dimensional Hand Force Capability Model for the Seated Operator" (University of Michigan, 1972); T Stobbe, "The Development of a Practical Strength Training Program for Industry" (University of Michigan, 1982).

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applicable variables in each case are used to estimate the mean maximum capacity and the standard deviation of the population distribution. The body build selected under **Model Settings** is then used to place the worker at a particular percentile of that distribution and obtain an individualized maximum capacity.

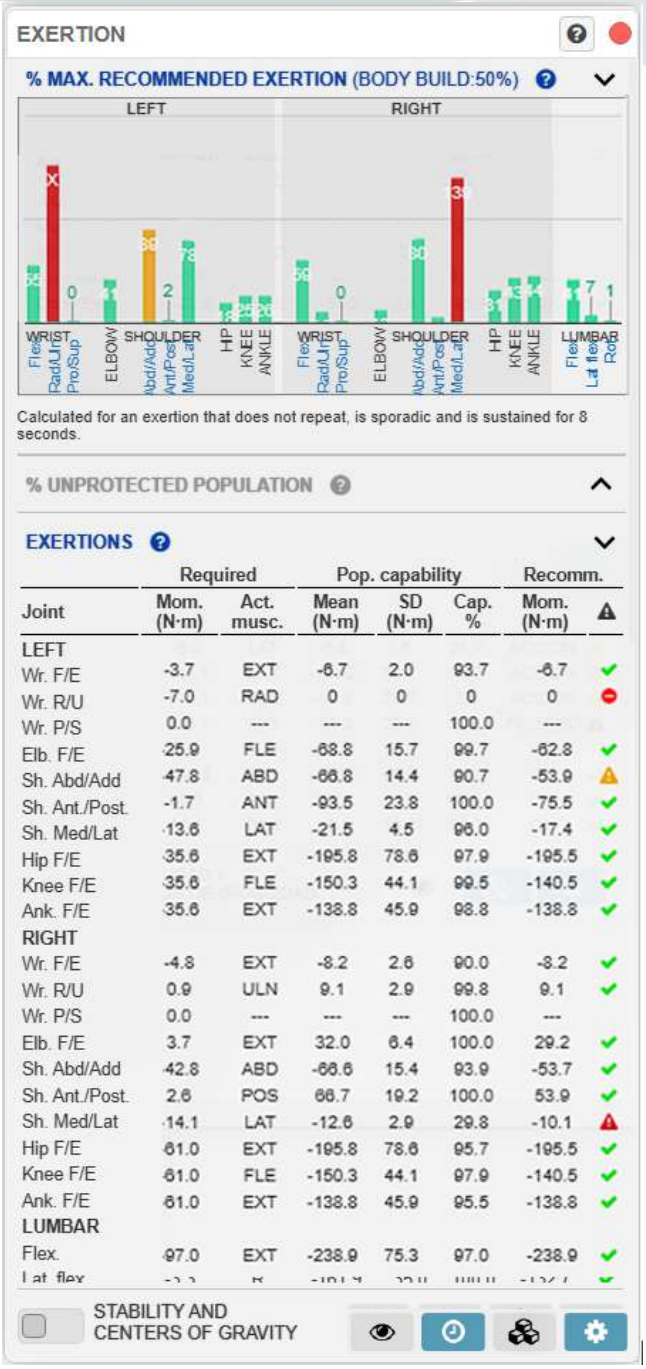
An example of this type of model is Equation 5, which shows how to calculate the mean maximum moment-generating capacity at the elbow joint when the active muscle group is the forearm flexor group. As the equation shows, the mean maximum capacity depends on the elbow flexion angle (α_{elbow}) and the shoulder flexion angle ($\alpha_{shoulder}$). It also depends on sex: *G* is the sex-related factor and takes different values for men and women. Factor *Z* accounts for whether the limb being assessed is on the person's dominant or nondominant side. Finally, *E* adjusts the mean maximum capacity based on age and sex.

$$M_{codo}^{flexión} = [336.29 + 1.544 * \alpha_{codo} - 0.0085 * \alpha_{codo}^2 - 0.5 * \alpha_{hombro}] \cdot G \cdot Z \cdot E$$

Equation 5

The value obtained with this type of model is the **mean maximum capacity** of the reference population. Because this quantity is assumed to follow an approximately normal distribution, the models also estimate its standard deviation. An individualized maximum capacity can be estimated from these two values and the percentile associated with the selected body build. This result is a statistical estimate, not a direct measurement of the assessed person's physical capacity.

When biomechanical analysis is enabled, Ergoniza Intelligent displays information on maximum capacity by joint, as shown in Figure 38 and Figure 40. The first column of the tables identifies the joint and type of exertion. The second column shows the component of the required joint moment associated with the movement being analyzed, expressed in a local reference system linked to the corresponding body segment. Therefore, its value does not necessarily match any of the X, Y, or Z components displayed in the **Joint loads** panel in Figure 37.



Joint	Required Mom. (N·m)	Required Act. musc.	Pop. capability Mean (N·m)	Pop. capability SD (N·m)	Pop. capability Cap. %	Recomm. Mom. (N·m)	
LEFT							
Wr. F/E	-3.7	EXT	-6.7	2.0	93.7	-6.7	✓
Wr. R/U	-7.0	RAD	0	0	0	0	⛔
Wr. P/S	0.0	---	---	---	100.0	---	
Elb. F/E	25.9	FLE	-68.8	15.7	99.7	-62.8	✓
Sh. Abd/Add	47.8	ABD	-66.8	14.4	90.7	-53.9	⚠
Sh. Ant./Post.	-1.7	ANT	-93.5	23.8	100.0	-75.5	✓
Sh. Med/Lat	13.6	LAT	-21.5	4.5	96.0	-17.4	✓
Hip F/E	35.6	EXT	-195.8	78.6	97.9	-195.5	✓
Knee F/E	35.6	FLE	-150.3	44.1	99.5	-140.5	✓
Ank. F/E	35.6	EXT	-138.8	45.9	98.8	-138.8	✓
RIGHT							
Wr. F/E	-4.8	EXT	-8.2	2.6	90.0	-8.2	✓
Wr. R/U	0.9	ULN	9.1	2.9	99.8	9.1	✓
Wr. P/S	0.0	---	---	---	100.0	---	
Elb. F/E	3.7	EXT	32.0	6.4	100.0	29.2	✓
Sh. Abd/Add	42.8	ABD	-66.6	15.4	93.9	-53.7	✓
Sh. Ant./Post.	2.6	POS	66.7	19.2	100.0	53.9	✓
Sh. Med/Lat	14.1	LAT	-12.6	2.9	29.8	-10.1	⛔
Hip F/E	61.0	EXT	-195.8	78.6	95.7	-195.5	✓
Knee F/E	61.0	FLE	-150.3	44.1	97.9	-140.5	✓
Ank. F/E	61.0	EXT	-138.8	45.9	95.5	-138.8	✓
LUMBAR							
Flex.	97.0	EXT	-238.9	75.3	97.0	-238.9	✓
Lat. flex.							

Figure 38: Exertion by joint as a percentage of maximum recommended exertion.

The third column of the tables identifies the functional muscle group that must act according to the direction of the required moment. Columns 4 through 6, grouped under the heading **Population capacity**, show the estimated maximum capacities for the current joint and posture. As noted above, these capacities vary among individuals and are assumed to be distributed approximately normally around the mean. The first of these columns shows the mean maximum capacity, and the second shows the standard deviation of the distribution. The last, **% protected population**, shows the estimated proportion

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of the population whose maximum capacity would be equal to or greater than the required moment. This percentage represents the proportion of the population with sufficient moment-generating capacity according to the model; it is not an individual's probability of injury.

For example, the first row of the table in Figure 38 indicates that the required flexion/extension moment at the left wrist is −3.7 N·m. Under the convention used in Ergoniza Intelligent, the negative sign identifies the extensor as the active muscle group. The estimated mean maximum capacity for that group is −6.7 N·m (a magnitude of 6.7 N·m), and the standard deviation is 2.0 N·m. Based on this distribution, the model estimates that 93.7% of the population would have sufficient capacity to generate the required moment.

10.1.1 Exertion assessment

The mean maximum capacity displayed by Ergoniza Intelligent in the exertion table represents the population mean of the maximum joint moment that each functional muscle group can generate under the analysis conditions. This quantity corresponds to the Maximum Voluntary Contraction estimated by the model.

An exertion close to Maximum Voluntary Contraction can be maintained only for a very short time and leaves little margin relative to the estimated maximum capacity. It is therefore not recommended as a normal condition of a work activity. Maximum joint capacity also varies among individuals and is assumed to be distributed approximately normally around the population mean. Consequently, exertion assessment must consider both the mean capacity and the dispersion of that distribution.

Ergoniza Intelligent provides two procedures for assessing joint exertions. The first adapts the assessment to the worker's characteristics and the task conditions. Within this procedure, the indicator name and the moment used as the reference depend on the analysis mode and the enabled settings. The result may be displayed as **Percentage of Recommended Joint Capacity**, **Percentage of Maximum Isometric Capacity Used** or **Percentage of Maximum Recommended Exertion**. The differences among these indicators are explained in the following sections.

The second procedure uses the distribution of joint capacities in the population and expresses the result as **Percentage of Protected Population**.

10.1.1.1 Customized exertion assessment based on worker and task characteristics

This procedure compares the required moment at each joint with a customized reference moment. The calculation starts from an estimate of the individualized maximum isometric capacity common to both analysis modes, but the method used to obtain the reference moment depends on whether a single posture or a multiple-posture sequence is being analyzed.

First, Ergoniza Intelligent uses the biomechanical equations described in the previous section to determine the mean maximum isometric capacity and the corresponding standard deviation for the active joint, posture, and muscle group. As discussed above, this calculation considers characteristics such as sex, age, and laterality or dominance.

From this population distribution, the body build defined under **Model Settings** is used to select a particular percentile and obtain the *individualized maximum isometric capacity*. Body-build values range from *Very slight* to *Very robust*: Very slight corresponds to the 5th percentile; Average, to the 50th percentile; and Very robust, to the 95th percentile.

The selected body build therefore modifies the maximum isometric capacity estimated for the worker, producing the individualized maximum isometric capacity. This individualized capacity is an estimate of the maximum moment-generating capacity and does not, by itself, constitute a recommended limit for performing a task.

In *Single Posture* mode, Ergoniza Intelligent determines the recommended *Percentage of Maximum Voluntary Contraction* (recommended %MVC) from the duration and repetitiveness of the exertion. This percentage indicates the fraction of the individualized maximum isometric capacity that is recommended under the task conditions. Its calculation procedure is explained in Section 10.2.

In this mode, the recommended %MVC is applied to the individualized maximum isometric capacity to obtain the *Base Maximum Moment* for each degree of freedom of each joint (z):

$$M_{base,z} = M_{isométrico,max,z} \cdot \frac{\%MVC_{recomendado,z}}{100}$$

Equation 6

For example, in *Single Posture* mode, if the individualized maximum isometric capacity is 100 N·m and, given the duration and repetitiveness of the task, no more than 15% MVC is recommended, the Base Maximum Moment will be 15 N·m. The Base Maximum Moment is expressed in N·m and constitutes the reference joint moment, already adjusted to

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the worker's body build and the temporal conditions of the analyzed action.

The criteria used to calculate the reference moment can be configured in the **Biomechanical Analysis Settings** dialog (Figure 41). This dialog is opened with the gear-shaped button at the bottom of the **Exertion** window. The **Calculate max. recommended exertion (%)** control enables or disables the epidemiological priority adjustment and is available in both *Single Posture* and *Multiple Postures* modes. In *Multiple Postures* mode, the dialog also displays the **Include cumulative exertion load in recommended exertion (%)** control, which enables or disables the temporal adjustment of the recommended moment.

In *Multiple Postures* mode, the reference moment is not obtained by applying the recommended %MVC for an isolated action to each posture. When **Include cumulative exertion load in recommended exertion (%)** is enabled, Ergoniza Intelligent uses the three-compartment model with enhanced recovery during rest, known as **3CC-r** (*Three-Compartment Controller with enhanced recovery*), an adaptation of the three-compartment model that represents fatigue and recovery during intermittent tasks[11]. This model represents the evolution of muscular capacity by distributing the estimated total capacity among three states: available capacity at rest, capacity activated to generate the required moment, and temporarily fatigued capacity. At each instant, the model can estimate what proportion of the individualized maximum isometric capacity remains available as recommended capacity. The resulting **3CC-r Recommended Moment** evolves according to the previous history of exertions, recovery, activity repetitions, and the rest periods between them.

If this control is disabled, the individualized maximum isometric capacity is used as the reference without a 3CC-r reduction. The same applies when the control is unavailable because **Include repetitions and rest periods** is disabled in the **Activity Repetitions** dialog (Figure 46). In *Multiple Postures* mode, this dialog can be opened with the **Timing** button on the side menu or the clock-shaped button at the bottom of the **Exertion** window (Figure 38).

Although the control name uses the term cumulative load, the 3CC-r adjustment does not use the value of the separate **Cumulative exertion load** indicator. That indicator is calculated separately using Maximum Endurance Time (MET) models and is explained in Section 10.3. The operation of the 3CC-r model and the relationship between the two procedures are described in Section 10.4.

Once the reference moment for the analysis mode has been determined, Ergoniza Intelligent compares it with the required moment to obtain the customized exertion indicator.

When **Calculate max. recommended exertion (%)** is disabled, the epidemiological priority adjustment is not applied. This adjustment weights the reference moment according to the observed prevalence of musculoskeletal disorders (MSDs) in the different body regions. The indicator displayed when the adjustment is disabled depends on the analysis mode and the moment used as the reference.

In *Single Posture* mode, the program displays the **Percentage of Recommended Joint Capacity** (%RJC), which compares the required moment with the Base Maximum Moment calculated from the duration and repetitiveness of the action.

In *Multiple Postures* mode, if **Include cumulative exertion load in recommended exertion (%)** is enabled, the program also displays the Percentage of Recommended Joint Capacity (%RJC), but in this case it compares the required moment with the 3CC-r Recommended Moment for the instant being analyzed. If this control is disabled or unavailable, the required moment is compared directly with the individualized maximum isometric capacity, and the result is called the **Percentage of Maximum Isometric Capacity Used**.

When **Calculate max. recommended exertion (%)** is enabled, the epidemiological priority adjustment is applied: the program weights the reference moment according to MSD prevalence in the body region and displays the **Percentage of Maximum Recommended Exertion** (%MRE), regardless of the analysis mode. Its calculation procedure is explained in Section 10.1.1.1.1.

In *Single Posture* mode, the Percentage of Recommended Joint Capacity (%RJC) expresses the required joint moment as a percentage of the Base Maximum Moment and is calculated for each degree of freedom of each joint (z) as follows:

$$\%CAR = 100 \cdot \frac{|M_{req,z}|}{|M_{base,z}|}$$

Equation 7

In this mode, a value of 100% indicates that the magnitude of the required moment equals the Base Maximum Moment;

[11] John M. Looft, Nicole Herkert, and Laura Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks," *Journal of Biomechanics* 77, no. 22 (2018): 16–25, doi:10.1016/j.jbiomech.2018.06.005.

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higher values indicate that it exceeds that moment. In *Multiple Postures* mode, when the 3CC-r adjustment is active, the equivalent ratio is established relative to the 3CC-r Recommended Moment, which may vary throughout the sequence. Its formulation is presented in Section 10.4.

10.1.1.1.1 Percentage of Maximum Recommended Exertion

The biomechanical indicator obtained without applying the epidemiological priority adjustment shows what proportion of the reference moment is demanded by the task. Depending on the analysis mode and the enabled settings, this indicator is called the **Percentage of Recommended Joint Capacity (%RJC)** or the **Percentage of Maximum Isometric Capacity Used**, as explained in the previous section.

The **Percentage of Maximum Recommended Exertion (%MRE)** supplements this assessment by incorporating the observed prevalence of musculoskeletal disorders in the different body regions. It thereby increases the priority assigned to regions whose prevalence is higher than the reference prevalence and reduces it in regions where prevalence is lower. The ergonomist thus has an indicator that combines the task's biomechanical demands with the observed epidemiological distribution of musculoskeletal disorders, helping prioritize analyses and possible intervention measures.

Body region	Prevalence
Lumbar region	47 %
Shoulder	50 %
Elbow	21 %
Wrist	42 %
Hip	11 %
Knee	33 %
Ankle	17 %

Table 2: Mean 12-month prevalences used in the epidemiological priority adjustment.

To enable this adjustment, select **Calculate max. recommended exertion (%)** under **Epidemiological priority adjustment** in the **Biomechanical Analysis Settings** dialog (Figure 41). This dialog is opened with the gear-shaped button at the bottom of the **Exertion window** (Figure 38).

In the instantaneous assessment, the epidemiological adjustment is applied after the reference moment for the analysis mode has been determined. It does not modify the required moment, the individualized maximum isometric capacity, the **Base Maximum Moment** or the internal evolution of the 3CC-r model. Its function is to weight the applicable reference moment and the percentage obtained from it.

The prevalences are taken from the systematic review and meta-analysis by Govaerts et al. (2021)[12] of workers in European secondary industries. The study's mean 12-month prevalences are used and assigned to the model regions as described below.

Ergoniza Intelligent internally defines a reference prevalence (P_{ref}) to normalize the adjustment. This reference is the arithmetic mean of the seven prevalences assigned to the model regions and has a value of 31.57%. For each body region z, the weighting factor is calculated as follows:

$$W_z = \left(\frac{P_z}{P_{ref}}\right)^{\lambda}$$

Equation 8

Here, P_z is the prevalence for the region and λ is the **Influence** parameter. The user can change this parameter using the **Influence** control under **Epidemiological priority adjustment** in the **Biomechanical Analysis Settings dialog** (Figure 41). It can take values from 0.1 to 0.9, and its initial value is 0.5. λ does not represent a percentage contribution; it controls the strength of the weighting. The higher its value, the greater the difference established between regions with high and low prevalences. If a region's prevalence equals the reference prevalence, its weighting factor is 1.

To express the calculation consistently in both analysis modes, $M_{ref,z}$ denotes the biomechanical reference moment obtained before applying the epidemiological adjustment. This moment corresponds to:

- The Base Maximum Moment in Single Posture mode.
- The 3CC-r Recommended Moment for the instant being analyzed in Multiple Postures mode when the 3CC-r adjustment is enabled.
- The individualized maximum isometric capacity in Multiple Postures mode when the 3CC-r adjustment is disabled or unavailable.

The reference limit resulting from the epidemiological adjustment is obtained by dividing the biomechanical reference moment by the weighting factor for the region:

[12] Renée Govaerts et al., "Prevalence and Incidence of Work-Related Musculoskeletal Disorders in Secondary Industries of 21st Century Europe: A Systematic Review and Meta-Analysis," *BMC Musculoskeletal Disorders* 22, no. 751 (2021), doi:10.1186/s12891-021-04615-9.

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$$M_{epi,z} = \frac{M_{ref,z}}{W_z}$$

Equation 9

The Percentage of Maximum Recommended Exertion is therefore calculated as follows:

$$\%EMR_z = 100 \cdot \frac{|M_{req,z}|}{|M_{epi,z}|}$$

Equation 10

This result is equivalent to multiplying the percentage that would be displayed without the epidemiological adjustment by W_z. A prevalence above the reference prevalence therefore increases the displayed percentage, while a prevalence below it reduces the percentage. This reduction does not mean that the joint has greater physiological capacity, that the required moment is lower, or that the task is biomechanically safer; it indicates only a lower relative epidemiological priority.

The %MRE should be interpreted as a comparative preventive-priority index, not as a measure of physiological capacity, an individual's probability of injury, or a causal estimate. The adjustment does not modify the mean maximum capacity, its standard deviation, or the **Percentage of Protected or Unprotected Population**. The same epidemiological weighting is applied to the Cumulative exertion load result, as explained in Section 10.3, which is presented as an indicator independent of the instantaneous %MRE and the temporal adjustment by 3CC-r.

When **Calculate max. recommended exertion (%)** is disabled, the displayed result depends on the mode and the reference moment used:

- In *Single Posture* mode, the **%RJC** calculated relative to the Base Maximum Moment is displayed.
- In *Multiple Postures* mode with the 3CC-r adjustment enabled, the **%RJC** calculated relative to the 3CC-r Recommended Moment is displayed.
- In *Multiple Postures* mode with the 3CC-r adjustment disabled or unavailable, the **Percentage of Maximum Isometric Capacity Used**, calculated relative to the individualized maximum isometric capacity, is displayed.

10.1.1.1.2 Representation in the interface

When the customized exertion-assessment procedure is used, Ergoniza Intelligent displays the required moment, the active muscle group, the population distribution of joint capacity, and the reference moment used to assess the task demands in the **Exertions** table in the **Exertion** window (Figure 38).

In the results table, the **Recomm**. column displays the percentage for the active assessment criterion. When the epidemiological priority adjustment is enabled, the **Percentage of Maximum Recommended Exertion** (%MRE) is displayed. When it is disabled, the **Percentage of Recommended Joint Capacity** (%RJC) is displayed if the reference is the Base Maximum Moment (in *Single Posture* mode) or the 3CC-r Recommended Moment (in *Multiple Postures* mode with this adjustment enabled). If the 3CC-r adjustment is not used in *Multiple Postures* mode, the percentage of maximum isometric capacity used is displayed.

The name of the indicator shown above the table and on the chart is updated automatically according to the analysis mode and enabled settings:

- **% maximum recommended exertion**, when the epidemiological priority adjustment is enabled.
- **% recommended joint capacity**, when the epidemiological adjustment is disabled and the reference moment incorporates the temporal adjustment for the analysis mode.
- **% maximum isometric capacity used**, in **Multiple Postures** mode when both the epidemiological adjustment and the 3CC-r adjustment are disabled.

The last column of the table shows the level reached by the calculated percentage. If the value is below the Warning threshold, a green check mark appears. When it reaches the Warning threshold, an orange exclamation mark appears; when it reaches the Risk threshold, the exclamation mark is red. The initial values of these thresholds are 85% and 100%, respectively.

The user can change these values with the **Orange threshold (%)** and **Red threshold (%)** controls in the **Visualization** panel of the settings side menu. The thresholds apply to the actual calculated percentage regardless of the name of the displayed indicator.

A red symbol with a horizontal white line indicates that the joint is in an extreme posture in which exertion is not recommended for the movement and direction being analyzed; the corresponding reference moment is therefore zero. This is the case for left-wrist lateral deviation in Figure 38. Situations involving passive joint structures are described in greater detail in the following section.

The chart at the top of the **Exertion** window represents the active indicator for the different joints and movements. Bars are green when the value is below the Warning threshold, orange when it lies between the Warning and Risk thresholds, and red when it reaches or exceeds the Risk threshold.

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When the 3CC-r temporal adjustment is used, the available recommended capacity may approach zero and produce very high percentages. To preserve readability, values above 500% are identified with the label **>500**. Bar height is visually limited to the chart's maximum scale; this limitation does not alter the actual value used to determine the risk level.

Figure 38 corresponds to an analysis performed in *Single Posture* mode. For this reason, the duration and repetitiveness conditions used to calculate the Base Maximum Moment are shown below the chart. In the example, the exertion is not repeated, is sporadic, and is sustained for 11 seconds. Left-shoulder abduction exceeds the Warning threshold, and right-shoulder lateral rotation greatly exceeds the Risk threshold. The worker's selected body build, corresponding to the 50th percentile in the example, is also shown above the chart.

In *Multiple Postures* mode, the percentage and the 3CC-r Recommended Moment may vary at each instant because of the previous history of exertions and recoveries. The displayed values therefore always correspond to the currently selected instant in the sequence.

10.1.1.1.3 Joint-limit situations at the elbow and knee

In certain postures, some joints—particularly those analyzed by the biomechanical model only through flexion and extension, such as the elbow and knee—may be at one end of their range of motion while the calculated external moment tends to move the joint beyond that limit, as in the case shown in Figure 39.

Under these circumstances, the joint moment still exists and must be resisted, but it should not be interpreted exclusively as a conventional active muscular demand. Part of the resistance may be borne by passive structures such as ligaments, the joint capsule, bone geometry, and periarticular tissues. Passive joint resistance is defined precisely as the torque exerted by viscoelastic tissues whose primary function is to restrict the range of motion[13].

When the elbow is extended, the anterior capsule plays an important role in resisting certain loads, and the collateral ligaments contribute to lateral stability when the elbow is extended[14]. At terminal knee extension or hyperextension, stability depends heavily on passive structures and joint locking, so a moment tending to extend the knee further should not automatically be interpreted as an extensor muscle demand.

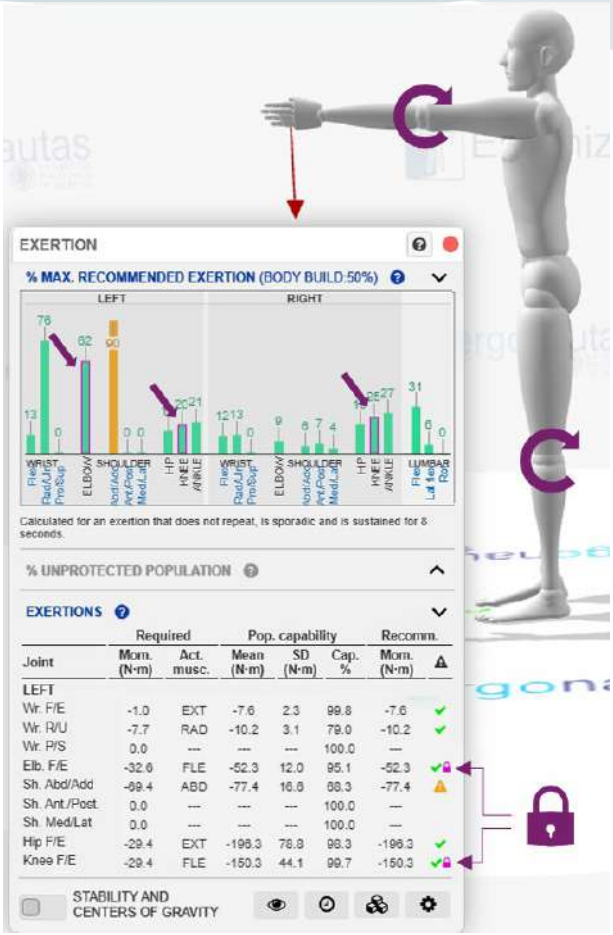


Figure 39: Identification of joint-limit situations at the elbow and knee.

For example, if the knee is fully extended and the external moment tends to produce further extension, the response is not necessarily equivalent to an extensor muscle contraction comparable with a percentage of maximum voluntary contraction (%MVC). In that case, the result should be interpreted as a joint-limit condition involving passive structures, not as an ordinary active muscular exertion.

In these situations, Ergoniza Intelligent continues to calculate and display the joint moment because the mechanical load on the joint remains relevant. However, when it detects this condition, it identifies it visually in the interface. As shown in Figure 39, the bars in the exertion chart retain the color corresponding to the calculated demand level but include a purple border to indicate that the result occurs in a joint-limit situation. In the exertion table, the same condition is indicated by a **padlock icon** next to the assessment indicator, warning the user that the value must be interpreted cautiously because the moment may include a substantial contribution from

[13] Kei AOKI and Nobutoshi YAMAZAKI, "The Role of Joint Resistance by Passive Tissues in Human Bipedal Walking," *Biomechanisms* 14 (1998): 59–68, doi:10.3951/biomechanisms.14.59.

[14] Saif Ul Islam et al., "The Anatomy and Biomechanics of the Elbow," *The Open Orthopaedics Journal* 39, no. 2 (2008): 141–54, doi:10.2174/1874325002014010095.

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passive structures and should not be interpreted directly as an ordinary active muscular demand.

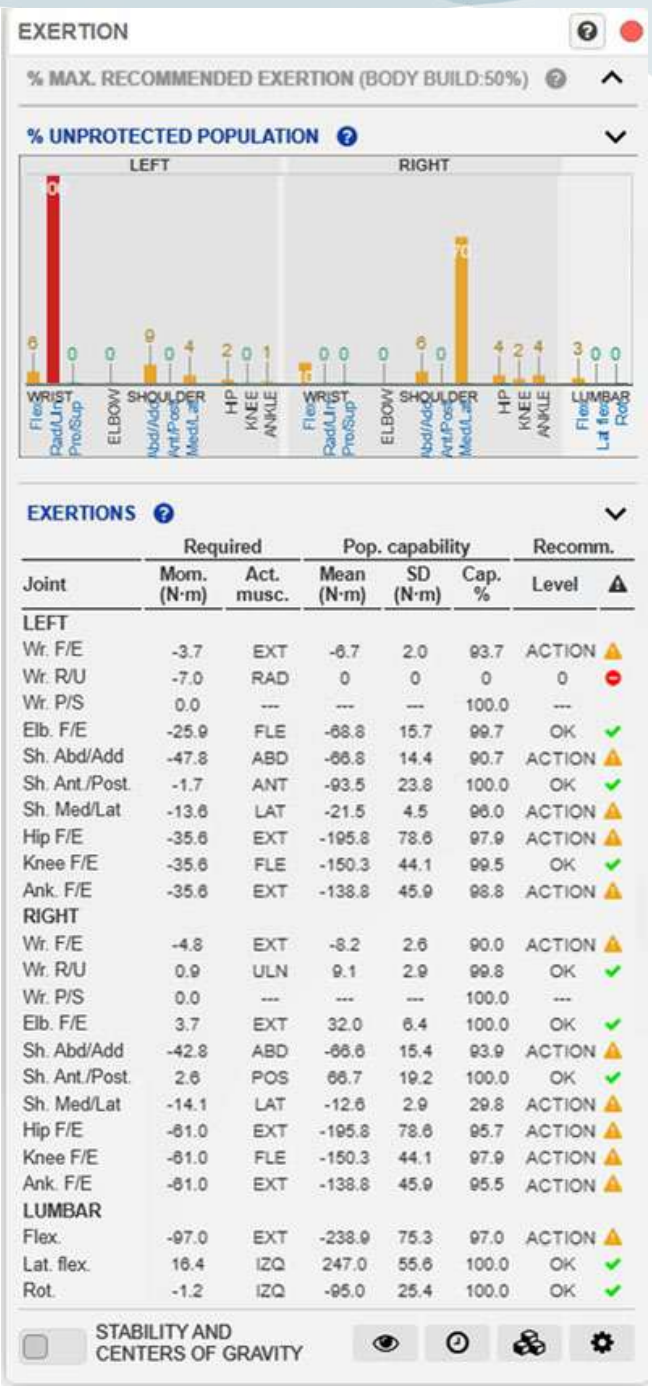
10.1.1.2 Exertion assessment by Percentage of Protected Population

The second exertion-assessment procedure estimates, for each joint and movement, the percentage of the reference population whose maximum capacity is equal to or greater than the required moment. Ergoniza Intelligent calls this result the *Percentage of Protected Population*, although in strictly biomechanical terms it represents the estimated percentage of the population with sufficient capacity to produce the analyzed moment. To classify the result, the program uses by default the action and danger limits derived from historical NIOSH criteria[15]. In this context, "protected population" does not mean a population free from risk; it means a population whose estimated capacity is not exceeded by the required moment.

Ergoniza Intelligent defines two different limits based on those established by NIOSH: the **Recommended Action Limit** and the **Danger Limit**. By default, the Recommended Action Limit is set at 99% of the male population capable and 75% of the female population capable. When the calculated percentage is below this limit, the result is classified at the Action level, indicating that the task conditions should be reviewed and possible corrective measures considered.

By default, the Danger Limit is set at 25% of the male population capable and 1% of the female population capable. When the calculated percentage is below this limit, the result is classified at the Danger level and requires priority preventive intervention.

The values used to define the Action and Danger limits can be customized in the settings for the **Exertion** window. To open them, click the gear-shaped button in the lower-right corner of the window, visible in Figure 40. The settings dialog is shown in Figure 41.



Joint	Required Mom. (N·m)	Required Act. musc.	Pop. capability Mean (N·m)	Pop. capability SD (N·m)	Pop. capability Cap. %	Recomm. Level
LEFT						
Wr. F/E	-3.7	EXT	-6.7	2.0	93.7	ACTION
Wr. R/U	-7.0	RAD	0	0	0	0
Wr. P/S	0.0	---	---	---	100.0	---
Elb. F/E	-25.9	FLE	-68.8	15.7	99.7	OK
Sh. Abd/Add	-47.8	ABD	-66.8	14.4	90.7	ACTION
Sh. Ant./Post.	-1.7	ANT	-93.5	23.8	100.0	OK
Sh. Med/Lat	-13.6	LAT	-21.5	4.5	96.0	ACTION
Hip F/E	-35.6	EXT	-195.8	78.6	97.9	ACTION
Knee F/E	-35.6	FLE	-150.3	44.1	99.5	OK
Ank. F/E	-35.6	EXT	-138.8	45.9	98.8	ACTION
RIGHT						
Wr. F/E	-4.8	EXT	-8.2	2.6	90.0	ACTION
Wr. R/U	0.9	ULN	9.1	2.9	99.8	OK
Wr. P/S	0.0	---	---	---	100.0	---
Elb. F/E	3.7	EXT	32.0	6.4	100.0	OK
Sh. Abd/Add	-42.8	ABD	-66.6	15.4	93.9	ACTION
Sh. Ant./Post.	2.6	POS	66.7	19.2	100.0	OK
Sh. Med/Lat	-14.1	LAT	-12.6	2.9	29.8	ACTION
Hip F/E	-61.0	EXT	-195.8	78.6	95.7	ACTION
Knee F/E	-61.0	FLE	-150.3	44.1	97.9	ACTION
Ank. F/E	-61.0	EXT	-138.8	45.9	95.5	ACTION
LUMBAR						
Flex.	-97.0	EXT	-238.9	75.3	97.0	ACTION
Lat. flex.	16.4	IZQ	247.0	55.6	100.0	OK
Rot.	-1.2	IZQ	-95.0	25.4	100.0	OK

Figure 40: Exertion by joint as a percentage of protected population.

[15] National Institute for Occupational Safety and Health (NIOSH), "Work Practices Guide for Manual Lifting -DHHS (NIOSH) Publication No. 81-122," *National Institute for Occupational Safety and Health (NIOSH)*, 1981.

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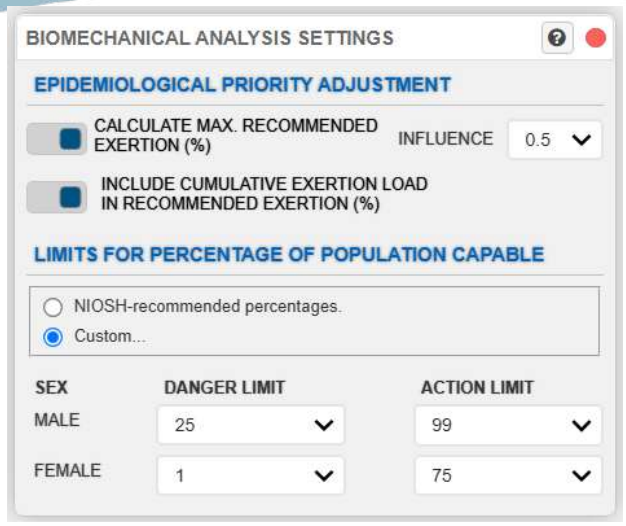


Figure 41: Biomechanical Analysis Settings in Multiple Postures mode.

When the Percentage of Protected Population criterion is selected in Ergoniza Intelligent, the final columns of the exertion table indicate whether the Recommended Action Limit or the Danger Limit has been reached. The upper chart in Figure 40 shows the percentage of unprotected population—the estimated percentage of the population whose joint capacity would be below the required moment. Assuming that the exertion is being performed by a man, all values above 1% reach the Recommended Action Limit; they are therefore shown in orange on the chart and marked with the word ACTION in the exertion table. For example, the chart in Figure 40 shows that 9% of the population has a joint capacity below the moment required for left-shoulder abduction, so the Recommended Action Limit is exceeded. If any value exceeded 75% of the unprotected population, it would exceed the Danger Limit and would be shown in red on the chart.

This procedure provides a population-based assessment rather than selecting a capacity corresponding to the body build of a particular worker. The capacity distributions used in the calculation incorporate the variables included in the biomechanical equations, such as posture, sex, age, and dominance when applicable. However, the procedure does not customize capacity using the body-build percentile or apply the reduction associated with task duration and repetitiveness. The default limits used to classify the result do differ for men and women.

In general, the Percentage of Protected Population criterion is particularly useful for assessing tasks intended for a large, heterogeneous working population because it indicates what proportion of that population is expected to have sufficient capacity to produce the required moments. When the aim is to analyze demands relative to a specific worker profile, the customized procedure is more appropriate. Depending on the analysis mode and the selected settings, the instantaneous indicator is expressed as the **Percentage of Recommended Joint Capacity (%RJC)**, **Percentage of Maximum Recommended Exertion (%MRE)** or **Percentage of Maximum Isometric Capacity Used**. In multiple-posture analysis, the task's temporal evolution is also assessed through the **Cumulative exertion load**.

10.2 Recommended Maximum Voluntary Contraction (%MVC) in a single-posture analysis

The **recommended Maximum Voluntary Contraction percentage** (recommended %MVC) is used in Ergoniza Intelligent to adapt the individualized maximum isometric capacity to the duration and repetitiveness conditions specified by the user when working in *Single Posture* mode. Applying this percentage yields the *Base Maximum Moment*, which is used as a reference for assessing the exertion required by the posture.

The procedure depends on whether the exertion is momentary, sustained static, or repetitive. The user defines these conditions in the **Repetitiveness** dialog shown in Figure 42. This dialog opens from the **Timing** button on the left sidebar or from the clock-shaped button at the bottom of the **Exertion** panel shown in Figure 38.

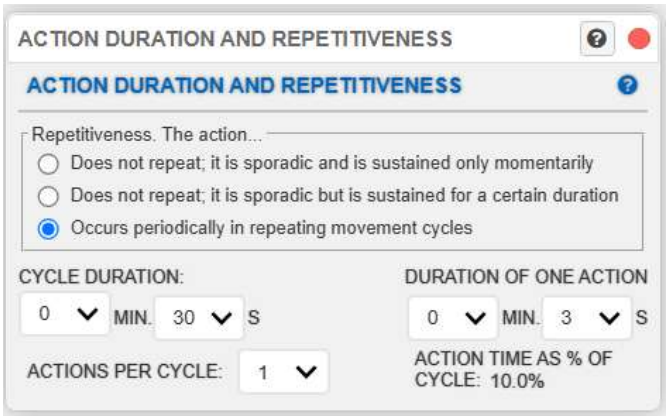


Figure 42: Action duration and repetitiveness settings.

In *Multiple Postures* mode, an independent recommended %MVC is not calculated for each posture in the sequence. This mode uses a specific procedure based on the time course of the activity and, where applicable, on the **3CC-r Recommended Moment** and cumulative load. The calculation of the 3CC-r Recommended Moment is explained in Section 10.4, while the separate procedure used to calculate cumulative exertion load is described in Section 10.3.

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The following sections describe only the calculation of the recommended %MVC in *Single Posture* mode for momentary exertions, sustained static exertions, and repetitive tasks.

10.2.1 Recommended %MVC for momentary exertions

When an exertion does not repeat, is sporadic, and is sustained only momentarily, Ergoniza Intelligent applies no additional reduction associated with duration or repetitiveness. The Base Maximum Moment therefore equals the individualized maximum isometric capacity for the joint and movement being analyzed. This capacity is obtained by considering the worker's posture, sex, age, laterality, and selected body build, as explained in Section 10.1.1.1.

Accordingly, a recommended %MVC of 100% is used for this type of exertion. This criterion should not be interpreted as a general recommendation to work continuously near maximum capacity, but as a reference applicable to momentary, sporadic actions of very short duration.

10.2.2 Recommended %MVC for static exertions

When an exertion is **static**, meaning that it is sustained over time, continuous muscle contraction can produce fatigue. Muscle fatigue can be defined as a reduction in the ability to generate force or power as a consequence of the activity performed, and it can contribute to the development of musculoskeletal disorders[16]. Various biomechanical models therefore incorporate procedures for estimating how fatigue develops[17], particularly in virtual human simulation applied to ergonomics[18]. Their purpose is to facilitate the assessment of sustained exertions and support the design of tasks and recovery periods.

Maximum Endurance Time (*MET*) is the maximum time for which a static contraction of a given intensity can be maintained before the exhaustion criterion is reached or the required exertion can no longer be sustained. MET decreases as contraction intensity increases and differs substantially among joint regions.

Muscle-fatigue modeling in ergonomics frequently uses Maximum Endurance Time models. These models make it possible to estimate the endurance of a muscle group or joint region without individually representing the activation and fatigue of each muscle. They are constructed from experimental data obtained for different contraction intensities and body regions. One of the best-known is Rohmert's model[19], traditionally used to analyze static contractions. This and other empirical models[20] provide reference values for limiting the intensity and duration of sustained exertions and for guiding the design of work and recovery periods.

For static exertions, Ergoniza Intelligent calculates the recommended %MVC using the joint-region-specific power models obtained by Frey Law and Avin[21] from a systematic review and meta-analysis of studies on static contractions.

These models relate relative contraction intensity to Maximum Endurance Time and show that this relationship varies among joint regions. Their general expression is presented in Equation 11, in which the parameters b_0 and b_1 depend on the joint region being analyzed (z).

$$METz = b_{0,z}\, MVC_z^{b1,z}$$

Equation 11

where:

- MET_z is the maximum endurance time, expressed in seconds.
- MVC_z is the contraction intensity expressed as a fraction of maximum capacity, between 0 and 1.
- $b_{0,z}$ and $b_{1,z}$ are the parameters corresponding to the joint region.

[16] R. H. Westgaard, "Work-Related Musculoskeletal Complaints: Some Ergonomics Challenges upon the Start of a New Century," *Applied Ergonomics* 31, no. 6 (2000): 569–80, doi:10.1016/S0003-6870(00)00036-3.
[17] L. Ma et al., "A Novel Approach for Determining Fatigue Resistances of Different Muscle Groups in Static Cases," *International Journal of Industrial Ergonomics* 41, no. 1 (2011): 10–18, doi:10.1016/j.ergon.2010.11.005; T. Xia and L. A. Frey Law, "A Theoretical Approach for Modeling Peripheral Muscle Fatigue and Recovery," *Journal of Biomechanics* 41, no. 14 (2008): 3046–52, doi:10.1016/j.jbiomech.2008.07.013.
[18] H. Honglun, S. Shouqian, and P. Yunhe, "Research on Virtual Human in Ergonomic Simulation," *Computers and Industrial Engineering* 53, no. 2 (2007): 350–56, doi:10.1016/j.cie.2007.06.027.
[19] W. Rohmert, "Ermittlung von Erholungspausen F¨ur Statische Arbeit Des Menschen," *European Journal of Applied Physiology* 18, no. 2 (1960): 123–164.
[20] K. El Ahrache, Da. Imbeau, and B. Farbos, "Percentile Values for Determining Maximum Endurance Times for Static Muscular Work," *International Journal of Industrial Ergonomics* 36, no. 2 (2006): 99–108, doi:10.1016/j.ergon.2005.08.003.
[21] Laura A.Frey Law and Keith G. Avin, "Endurance Time Is Joint-Specific: A Modelling and Meta-Analysis Investigation," *Ergonomics* 53, no. 1 (2010): 109–29, doi:10.1080/00140130903389068.

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In Ergoniza Intelligent, the user can use the dialog shown in Figure 42 to set the exertion duration after selecting "*Does not repeat; it is sporadic but is sustained for a certain duration*" in the Repetitiveness dialog. Given the entered exertion duration, Ergoniza Intelligent calculates the recommended maximum voluntary contraction by solving Equation 11 for MVC.

As a preventive criterion, Ergoniza Intelligent requires the entered duration not to exceed 50% of the maximum endurance time estimated by the models, [22]7 (American Conference of Governmental Industrial Hygienists). This introduces a margin relative to the experimentally predicted time to exhaustion. The criterion is used to obtain a recommended contraction intensity and should not be interpreted as an exact prediction of when fatigue will occur in a particular worker. ACGIH recommends using this threshold as a practical reference for limiting the duration of static muscle contractions.

10.2.3 Recommended %MVC for repetitive tasks

Repetitive tasks involve performing similar movements or exertions cyclically. These tasks are organized into work cycles that repeat continuously. Repetitiveness means that the same muscle groups are activated repeatedly within each cycle, which can lead to fatigue accumulation and increase the risk of developing musculoskeletal disorders if exertion and recovery times are not managed appropriately.

When an Ergoniza Intelligent user selects "*Occurs periodically in repeating movement cycles*" in the Repetitiveness dialog shown in Figure 42, the user must enter the work-cycle duration, the duration of one exertion, and the number of exertions performed in each work cycle. Using these data, Ergoniza Intelligent calculates the percentage of the total work-cycle duration occupied by exertion.

Once this value has been obtained, the procedure established by ACGIH (2016) is used to calculate the recommended maximum voluntary contraction percentage (%MVC) for repetitive work tasks, that is, tasks performed in activity cycles. The ACGIH TLV curve is based primarily on psychophysical data concerning maximum acceptable exertions, supplemented by physiological and biomechanical evidence. Its purpose is to establish recommended combinations of intensity and duty cycle to limit the development of persistent localized fatigue in the upper limbs.

The *Threshold Limit Value* (TLV) curve defines the recommended %MVC as a function of the *duty cycle*, DC, meaning the percentage of the total cycle time during which muscular exertion occurs. In the ACGIH procedure, activity time is the time during which the exertion exceeds 5% of maximum capacity. In Ergoniza Intelligent, the user identifies this time by entering the duration of each exertion and the number of exertions performed during the cycle; the program uses these data to calculate DC with Equation 12.

$$DC = 100 \cdot \frac{n \cdot t_e}{T_c}$$

Equation 12

where:

- n is the number of exertions per cycle,
- t_e is the duration of each exertion, and
- T_c is the total cycle duration.

If the actual %MVC exceeds the value recommended by the TLV curve for a given DC, this indicates an unfavorable condition according to the localized-fatigue criterion and suggests that the exertion intensity, duration, or frequency should be reduced.

To determine this limit quantitatively, Equation 13 is used.

$$\%MVC_{TLV} = 100 \cdot \left[-0.143 \cdot ln\left(\frac{DC}{100}\right) + 0.066\right]$$

Equation 13

where $\%MVC_{TLV}$ is the maximum percentage recommended by the curve for the work cycle being analyzed, and DC (duty cycle) is the portion of the cycle time during which the exertion is performed, expressed as a percentage.

To obtain the final recommended %MVC, Ergoniza Intelligent multiplies the value provided by the TLV curve by the adaptation coefficient for the anatomical region, shown in Table 3. In all cases, the result is limited to 100% of the individualized maximum isometric capacity.

The ACGIH TLV curve is defined specifically for localized upper-limb fatigue, considering exertions involving the hand/wrist, forearm, elbow, and shoulder. Its direct application to the trunk and lower limbs therefore cannot be considered a normative equivalence.

[22] ACGIH, "Ergonomic Threshold Limit Value (TLV) for Hand Activity Level (HAL) and Muscle Fatigue," in *2016 TLVs and BEIs* (Cincinnati, OH: ACGIH, 2016), ERG-1–ERG-14.

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For the trunk and lower limbs, there is no general-purpose TLV curve directly comparable to the one proposed by ACGIH for the upper limb. Nevertheless, the literature on maximum endurance times shows that fatigue does not develop in the same way in every joint and that endurance time depends on the body region involved. In particular, published endurance-time models show systematic differences among joints, with greater relative endurance at the ankle and lower endurance in regions such as the shoulder and knee.

Anatomical region	Coeff.	Rationale
Upper limb	1.0	Direct application of the ACGIH TLV curve.
Trunk	1.0	No increase, based on a preventive criterion and the absence of an equivalent general TLV for the trunk.
Hip / thigh	1.3	Moderate adaptation in the absence of a specific TLV.
Knee	1.2	Moderate adaptation; the literature does not support a generally high tolerance.
Ankle	1.8	Greater adaptation because of the higher relative fatigue resistance described in endurance-time models.

Table 3: Internal adaptation coefficients applied to the ACGIH TLV curve to estimate the recommended %MVC for repetitive tasks.

For this reason, Ergoniza Intelligent uses the ACGIH TLV curve as a general reference for representing the effect of the *duty cycle* on the maximum recommended voluntary contraction percentage, but introduces the internal adaptation coefficients shown in Table 3 to account for relative endurance differences among anatomical regions.

For this reason, Ergoniza Intelligent uses the ACGIH TLV curve as a general reference for representing the effect of the *duty cycle* on the maximum recommended voluntary contraction percentage, but introduces an internal adaptation coefficient based on the anatomical region (Table 3). This coefficient should not be interpreted as a specific TLV for the trunk or lower limbs, but as an ergonomic approximation intended to prevent a curve defined for the upper limb from being applied indiscriminately to the entire body, consistent with the greater relative fatigue resistance described for some joints in maximum endurance-time models (*endurance time*) [23].

10.3 Cumulative exertion load in multiple-posture analysis

When the activity being analyzed consists of a sequence of postures, exertion assessment cannot be based solely on the maximum instantaneous value reached at a joint or on the maximum voluntary contraction percentage recommended for an isolated posture. For a task performed over time, the intensity and duration of joint moments, their repetition, alternation between functional muscle groups, and the partial recovery that may occur during periods of lower demand or inactivity must also be considered.

For this reason, Ergoniza Intelligent incorporates a procedure for calculating cumulative exertion load in posture sequences. Its purpose is to estimate, for each joint and biomechanical channel, the cumulative exposure resulting from the time course of relative exertion. Unlike single-posture analysis, in which the Base Maximum Moment is determined from the duration or repetitiveness conditions defined by the user, multiple-posture analysis processes the complete time course of the activity.

At each time interval, the program calculates the relative exertion intensity, transforms it into a temporal dose, and integrates it while accounting for partial recovery. Doses are kept separate by joint, biomechanical channel, and functional muscle group, so that exertions corresponding to antagonist groups are not added as though they constituted the same muscular exposure.

The result should be interpreted as a cumulative exertion load or cumulative biomechanical dose. It does not represent the instantaneous maximum voluntary contraction percentage, a direct probability of injury, or an individual physiological measurement of muscle fatigue. The following sections describe the basis of the procedure, its calculation stages, and the criteria used to interpret its results.

10.3.1 Concept of cumulative exertion load

A task may involve moderate instantaneous exertions yet generate relevant biomechanical exposure if those exertions are sustained for a prolonged period, repeated frequently, or repeatedly engage the same functional muscle group. Conversely, a high but very brief exertion may have a different significance than a similar exertion that is sustained or repeated without sufficient recovery periods.

[23] Law and Avin, "Endurance Time Is Joint-Specific: A Modelling and Meta-Analysis Investigation."

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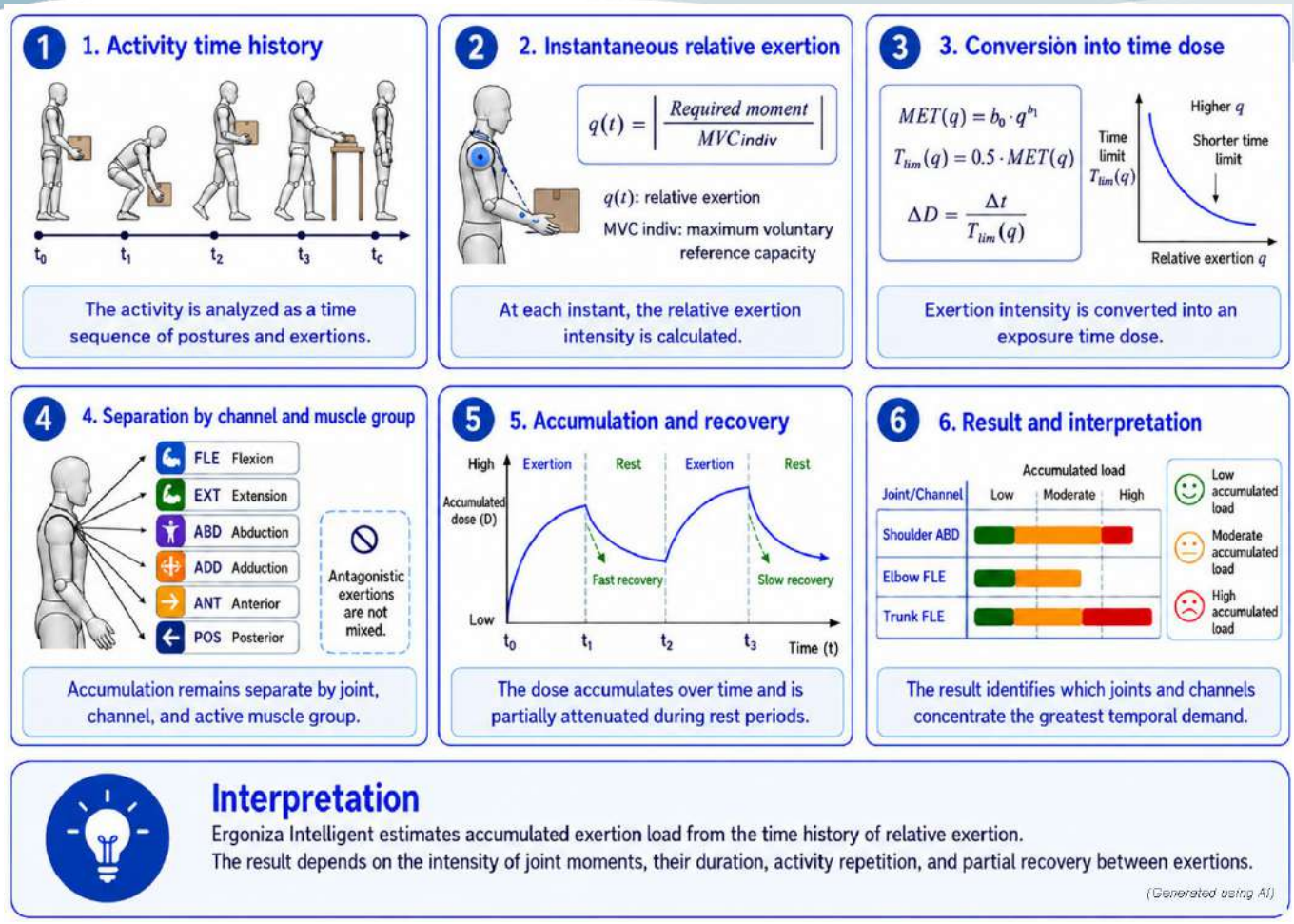


Figure 43: Calculation of cumulative exertion load in multiple postures.

For this reason, in multiple-posture analysis, Ergoniza Intelligent incorporates the concept of **cumulative exertion load** (Figure 43). This quantity represents the temporal biomechanical exposure resulting from the history of relative exertions produced during the activity. Its calculation is not intended merely to identify the most unfavorable instant in the sequence, but to assess how the combination of intensity, duration, repetition, and partial recovery determines the cumulative demand on the joints and functional muscle groups involved.

The starting point for the calculation is the relative exertion required at each instant in the sequence. This is obtained by relating the joint moment needed to maintain the posture and balance external loads to the individualized maximum isometric capacity corresponding to the biomechanical channel and active functional muscle group. Expressing exertion in this form makes it possible to compare demands of different magnitudes relative to the functional capacity estimated for each channel and posture. However, instantaneous relative exertion is not itself the result of the cumulative analysis; it is the input variable from which the temporal exertion dose is calculated.

The transformation of relative exertion into a temporal dose is based on the nonlinear relationship between the relative intensity of a contraction and the time for which it can be maintained before fatigue is reached. This relationship has been described using *maximum endurance time* (MET) models, including the joint-region-specific models developed by Frey Law and Avin[24]. These models show that, as relative exertion increases, the time for which the contraction can be maintained decreases sharply.

Under this approach, each interval in the sequence is interpreted as consuming a fraction of the time limit corresponding to the observed exertion level. The procedure therefore does not directly add joint moments or maximum voluntary contraction percentages, but rather temporal exposure doses. A given interval makes a greater contribution

[24] Ibid.

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when relative exertion is high and a smaller contribution when it is low. The resulting quantity can therefore be understood as a cumulative biomechanical dose associated with exertion.

Accumulation is performed separately for each joint, biomechanical channel, and functional muscle group. In this context, a muscle group represents the functional group that must act to resist the calculated moment (for example, flexors, extensors, abductors, or adductors), not an individual anatomical muscle. Exertions corresponding to antagonist groups are kept separate because they do not represent the same muscular exposure. Thus, a moment requiring flexor action is not accumulated together with another that engages the extensors, and the same approach is used for the remaining exertion directions.

The procedure also recognizes that previous exposure does not retain the same relevance indefinitely. When the demand on a muscle group decreases or ceases, partial recovery may occur. This recovery is progressive and depends on the time elapsed under reduced demand or at rest. The treatment of this phenomenon is based on the dynamic fatigue and recovery models proposed by Ma et al.[25] and by Xia and Frey Law[26], as well as later developments by Frey Law et al.[27], Looft et al.[28], and Sonne and Potvin[29] for sustained exertions, intermittent tasks, and complex force-time histories.

Cumulative exertion load should be interpreted as an ergonomic indicator of temporal biomechanical exposure. Its purpose is to help identify which joints and muscle groups concentrate the greatest demand over the course of an activity, assess the effect of repetition, estimate the influence of rest periods, and compare alternative task designs or performance methods. It is not equivalent to the instantaneous maximum voluntary contraction percentage, an average of exertions, or a direct prediction of injury or individual physiological fatigue. Its value lies in providing a structured measure of how relative muscular demand is distributed and accumulated during a posture sequence, within the biomechanical and ergonomic assumptions adopted for the analysis[30].

10.3.2 Instantaneous relative exertion

As explained in the description of customized exertion assessment (Section 10.1.1.1) and the maximum voluntary contraction percentage recommended for an isolated posture (Section 10.2), the demand at a joint can be expressed in relative terms by comparing the required joint moment with the maximum isometric capacity estimated for the biomechanical channel and active functional muscle group. In multiple-posture analysis, this same logic is applied successively throughout the entire time sequence.

For each biomechanical channel c, instantaneous relative exertion is defined as:

$$q_c(t) = \left| \frac{M_{req,c}(t)}{M_{isom,max,c,m}(t)} \right|$$

Equation 14

where:

- $M_{req,c}(t)$ is the joint moment required in biomechanical channel *c* at time *t*,
- $M_{isom,max,c,m}(t)$ is the individualized maximum isometric moment estimated for that same channel, posture, and functional muscle group *m*, and
- *m* is the functional muscle group that must act to resist the moment required at that time.

The direction of the required moment determines the active functional muscle group. For example, depending on the sign convention established for each channel, one moment direction may require flexor action and the opposite direction may require extensor action. Once the muscle group has been identified, relative exertion is calculated using the absolute value of the ratio, because the quantity relevant to determining the temporal dose is the proportion of its maximum capacity that the group must develop.

The value $q_c(t)$ is dimensionless and represents the fraction of maximum isometric capacity required at that time. Thus, a value of 0.25 indicates that the required moment equals 25%

[25] Liang Ma et al., "A New Simple Dynamic Muscle Fatigue Model and Its Validation," *International Journal of Industrial Ergonomics* 39 (2009): 211–20, doi:10.1016/j.ergon.2008.04.004.
[26] Xia and Frey Law, "A Theoretical Approach for Modeling Peripheral Muscle Fatigue and Recovery."
[27] Laura A. Frey-Law, John M. Looft, and Jesse Heitsman, "A Three-Compartment Muscle Fatigue Model Accurately Predicts Joint-Specific Maximum Endurance Times for Sustained Isometric Tasks," *Journal of Biomechanics* 45 (2012): 1803–8, doi:10.1016/j.jbiomech.2012.04.018.
[28] Looft, Herkert, and Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks."
[29] Michael W. Sonne and Jim R. Potvin, "A Modified Version of the Three-Compartment Model to Predict Fatigue during Submaximal Tasks with Complex Force-Time Histories," *Ergonomics* 59, no. 1 (2016): 85–98, doi:10.1080/00140139.2015.1051597.
[30] H. Iridiastadi and M. A. Nussbaum, "Muscle Fatigue and Endurance during Repetitive Intermittent Static Efforts: Development of Prediction Models," *Ergonomics*, 2006, doi:10.1080/00140130500475666.

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of the maximum capacity estimated for the corresponding channel and muscle group. A value of 1 equals 100% of that capacity, while a value greater than 1 indicates that the required moment exceeds the estimated maximum isometric capacity.

This calculation uses the individualized maximum isometric capacity without applying the reduction defined for the duration or repetitiveness of an isolated posture. The Base Maximum Moment calculated in single-posture analysis is therefore not used as the denominator. In the cumulative analysis, the influence of time is introduced later by transforming instantaneous relative exertion into a temporal exposure dose. This avoids counting the effects of duration and repetitiveness twice.

This normalization means that the procedure does not directly accumulate joint moments expressed in newton-meters (N·m), but rather exertions relative to estimated functional capacity. Consequently, the same required moment may represent different relative intensities depending on the joint, the adopted posture, and the functional muscle group that must generate it.

When the demand in a channel is resolved by a passive joint limit rather than the action of a muscle group, that interval does not add a new ordinary muscular dose. However, the previously accumulated load continues to decrease according to the partial-recovery procedure described in Section 10.3.7.

10.3.3 Determining the exposure time limit using MET curves

Once instantaneous relative exertion $q_c(t)$ has been determined, the next step is to relate it to the length of time for which it could be sustained continuously. For this purpose, models of *maximum endurance time* (MET) are used. These models express the relationship between the relative intensity of a contraction and the maximum time for which it can be sustained.

As explained for static exertions, this relationship is nonlinear. As relative exertion increases, maximum endurance time decreases rapidly. Small differences in the required relative exertion can therefore produce substantial differences in the time for which it could be sustained. This behavior has been described in the classic literature on static contractions and in later endurance-time models, including joint-specific models[31].

Ergoniza Intelligent uses the power models obtained by Frey Law and Avin from a meta-analysis of endurance times observed in different body regions. For body region z corresponding to the biomechanical channel being analyzed, maximum endurance time is calculated as follows:

$$MET_z(q) = b_{0,z} \cdot q^{b_{1,z}}$$

Equation 15

where $MET_z(q)$ is the estimated maximum endurance time, expressed in seconds, for body region z; q is relative exertion expressed as a fraction of maximum isometric capacity; and $b_{0,z}$ and $b_{1,z}$ are the coefficients for that region (Table 4). In all the models used, exponent $b_{1,z}$ is negative, so maximum endurance time decreases as relative exertion intensity increases.

Region	$b_{0,z}$	$b_{1,z}$
Wrist	33.55	−1.61
Elbow	17.98	−2.21
Shoulder	14.86	−1.83
Trunk	22.69	−2.27
Knee	19.38	−1.88
Ankle	34.71	−2.06
Hip	21.92	−1.98

Table 4: Coefficients of the Maximum Endurance Time models

The specific Frey Law and Avin models cover the ankle, trunk, hand or grip, elbow, knee, and shoulder. Because the study does not provide a specific model for the hip, Ergoniza Intelligent uses the general model obtained from the complete data set for that region. The hand or grip model is used for wrist exertions.

The value $MET_z(q)$ represents an estimate of the maximum time for which the exertion can be sustained until fatigue. However, using maximum endurance time as a direct operational limit is not appropriate in ergonomic analysis because that value is associated with a limiting condition of exhaustion. Therefore, Ergoniza Intelligent applies a preventive criterion and defines the exposure time limit as 50% of the estimated maximum endurance time:

$$T_{lim,z}(q) = 0.5 \cdot MET_z(q)$$

Equation 16

The factor 0.5 is an internal preventive criterion in Ergoniza Intelligent and establishes a reference time below the

[31] Rohmert, "Ermittlung von Erholungspausen F¨ur Statische Arbeit Des Menschen"; El Ahrache, Imbeau, and Farbos, "Percentile Values for Determining Maximum Endurance Times for Static Muscular Work"; Law and Avin, "Endurance Time Is Joint-Specific: A Modelling and Meta-Analysis Investigation."

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estimated maximum time to fatigue. It should not be interpreted as a regulatory limit or as a universal guarantee that fatigue will not occur. Its use is conceptually consistent with the preventive approach of the ACGIH limit values[32] for localized muscle fatigue, in which exhaustion is not used as the direct reference for acceptability.

The time limit $T_{lim,z}(q)$ should be interpreted as the reference exposure time for the relative exertion level observed at a given instant. If relative exertion is low, the time limit will be long and the cumulative contribution of a brief time interval will be small. Conversely, if relative exertion is high, the time limit will be short and the same time interval will consume a larger fraction of the allowable time.

Transforming $q_z(t)$ into $T_{lim,z}(q)$ converts an instantaneous measure of relative muscle intensity into a temporal quantity. This time limit can then be used to calculate, for each interval in the sequence, the fraction of the preventive reference time limit that has been consumed. That fraction constitutes the temporal exertion dose that will subsequently be integrated in the cumulative-load calculation.

10.3.4 Calculating the temporal exertion dose

Once the time limit $T_{lim,z(c)}(q_c(t))$ corresponding to the relative exertion observed in a biomechanical channel has been obtained, each time interval in the activity can be transformed into a fraction of the reference time limit. This fraction is called the **temporal exertion dose** and represents the portion of the time limit corresponding to the exposure produced during an interval of the activity.

For a biomechanical channel c, the elementary temporal dose is calculated as follows:

$$\Delta D_c(t) = \frac{\Delta t}{T_{lim,z(c)}(q_c(t))}$$

Equation 17

where:

- Δt is the duration of the time interval being considered, expressed in seconds,
- $q_c(t)$ is the relative exertion in channel c,
- z(c) is the body region to which the biomechanical channel belongs, and
- $T_{lim,z}(q_c(t))$ is the exposure time limit corresponding to that region and the relative exertion being considered.

This formulation makes it possible to interpret exposure in proportion to the time limit corresponding to the exertion intensity. For example, if the preventive time limit for a given relative exertion were 100 seconds, sustaining that exertion for 10 seconds would represent a temporal dose of 0.10. If, by contrast, relative exertion were higher and the time limit were reduced to 20 seconds, those same 10 seconds would represent a dose of 0.50. The same chronological duration can therefore make a very different cumulative contribution depending on the relative intensity of the exertion.

To prevent very low exertions from generating numerically insignificant doses, Ergoniza Intelligent applies a minimum threshold of 5% of the individualized maximum isometric capacity. When $q_c(t)$ does not exceed 0.05, the interval adds no new dose, although the existing cumulative load continues to decrease through recovery. When calculating dose with MET curves, relative exertions above 1.20 are evaluated as 1.20, that is, as 120% of maximum isometric capacity. This cap prevents the MET curves from being extrapolated beyond the interval adopted for the calculation; it does not modify the required moment or the instantaneous indicator displayed by the program.

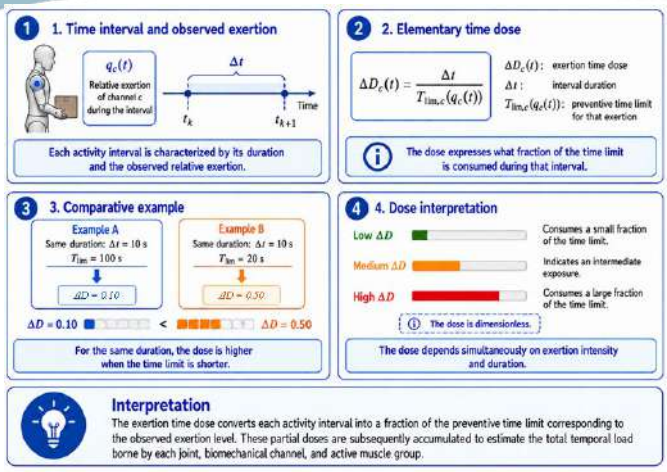
The dose is not calculated directly from the joint moment expressed in newton-meters (N·m), but from the moment normalized to the individualized maximum isometric capacity. Consequently, a moderate moment at a joint with lower relative capacity may generate a larger dose than a mechanically greater moment at another joint with higher capacity.

A sequence may contain brief intervals of high exertion, prolonged periods of moderate exertion, and periods of low or zero demand. Each interval whose exertion exceeds the threshold contributes a dose proportional to its duration and inversely proportional to its corresponding time limit. Thus, the result depends not only on the instantaneous maximum, but on the entire time history of the exertion.

The elementary temporal dose is dimensionless. A value of ΔD_c = 0.20 indicates that the interval being considered has consumed 20% of the preventive time limit associated with the observed relative exertion. Doses generated in successive intervals are integrated while accounting for partial recovery to estimate the cumulative load of each biomechanical channel and functional muscle group during the activity.

[32] ACGIH, "Ergonomic Threshold Limit Value (TLV) for Hand Activity Level (HAL) and Muscle Fatigue."

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(Generated using AI)

Figure 44: Calculation of the temporal exertion dose

This approach (Figure 44) makes it possible to compare exposures with different time profiles. Two tasks with the same maximum instantaneous exertion may generate different cumulative doses if one sustains the exertion longer, repeats it more frequently, or provides less recovery between exertions. Similarly, a task without very high peaks may reach a relevant cumulative load if it sustains a moderate demand for prolonged periods.

10.3.5 Separation by joint, biomechanical channel, and functional muscle group

The temporal exertion dose is not accumulated globally for the entire body or indiscriminately for a whole joint. A single joint may generate moments about different movement axes and, within each axis, may alternately require the action of antagonist functional muscle groups. The calculation is therefore organized at three levels: joint, biomechanical channel, and functional muscle group.

The **joint** identifies the anatomical location of the exposure. Left and right joints are calculated separately where applicable because they may adopt different postures and be subjected to different loads.

A **biomechanical channel** represents a degree of freedom or functional axis of moment generation. For example, flexion-extension constitutes one channel, abduction-adduction another, and rotations about another axis may constitute a third channel. The two opposite directions of the same movement are therefore not considered different channels, but antagonist actions within the same channel.

This distinction is consistent with the way moments and maximum isometric capacities are calculated. A joint may have different capacities depending on the movement axis and the functional muscle group that must act. The demand corresponding to flexion-extension should therefore not be combined with that corresponding to abduction-adduction or rotation.

Within each channel, **functional muscle groups** are distinguished. A muscle group does not represent an individual anatomical muscle, but the functional group that must generate the resisting moment: for example, flexors or extensors, abductors or adductors, and rotators in either direction.

The direction of the required moment, according to the convention established for each channel (see Section 8), identifies the active muscle group at each instant. Once it has been identified, the magnitude of relative exertion is obtained from the absolute value of the ratio defined above, and the corresponding temporal dose is assigned exclusively to that muscle group.

This separation is necessary because antagonist exertions should not accumulate as though they constituted continuous exposure of the same functional group. For example, a phase in which the flexors act and another in which the extensors act belong to the same flexion-extension channel, but generate doses in different muscle groups. When the exertion changes direction, the new dose is added to the group that becomes active, while the cumulative history of the antagonist group is retained separately and may decrease through recovery.

Conceptually, the cumulative load corresponding to joint *a*, biomechanical channel *c*, and functional muscle group *m* is represented by $D_{a,c,m}(t)$. Each combination of joint, channel, and muscle group maintains its own time course. When a channel contains several antagonist muscle groups, its summary value is not obtained by adding their loads. Instead, the greatest cumulative load among the functional muscle groups in the channel is selected:

$$D_{a,c}(t) = \max_{m \in M_{a,c}} D_{a,c,m}(t)$$

Equation 18

where $M_{a,c}$ is the set of functional muscle groups associated with channel *c* of joint *a*. Thus, if the flexors have a cumulative load of 0.60 and the extensors a load of 0.40, the summary value for the flexion-extension channel is 0.60. The group that determines the result may change over the course of the activity as the doses and recovery evolve.

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10.3.6 Temporal integration of the dose and cumulative-load update

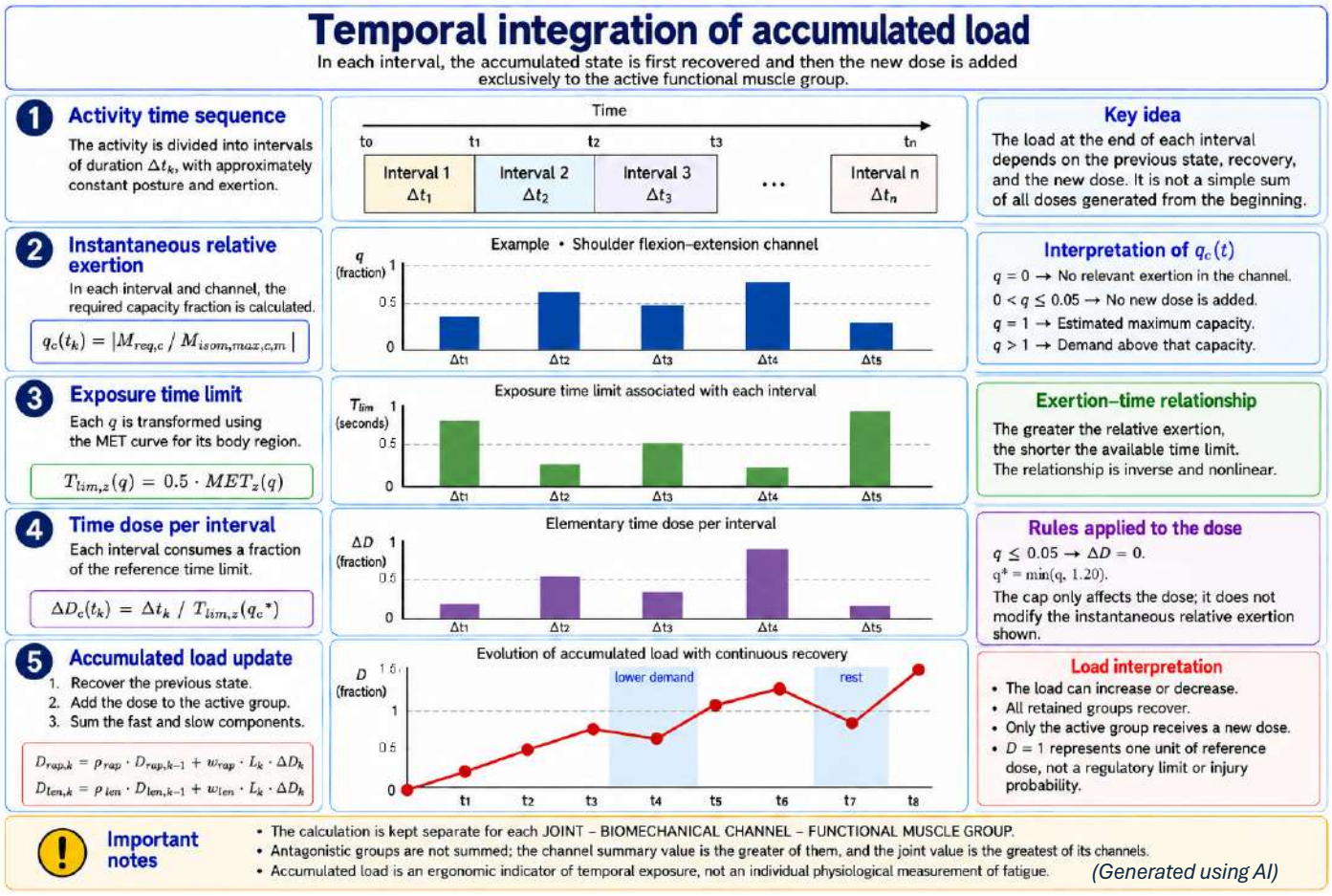


Figure 45: Temporal integration of cumulative load.

Once the elementary temporal dose for each interval has been calculated, the cumulative exertion load is obtained by processing all activity intervals in sequence. Each interval is characterized by its duration, the relative exertion in each biomechanical channel, and the functional muscle group that must act.

Integration does not consist solely of adding all the doses generated since the beginning of the activity. Previously accumulated exposure loses some of its value through recovery, including during intervals in which a new dose is generated. Two operations are therefore performed successively in each interval:

- Recovery corresponding to the interval duration is applied to the load previously accumulated by all muscle groups.
- The new temporal dose is added exclusively to the active functional muscle group.

To represent this process, the cumulative load of each muscle group is divided into a fast-recovery load component and a slow-recovery load component. For joint *a*, biomechanical channel *c*, functional muscle group *m*, and interval *k*, both components are updated as follows:

$$D^{rap}_{a,c,m,k} = \rho_{rap}(\Delta t_k) D^{rap}_{a,c,m,k-1} + \omega_{rap} I_{a,c,m,k} \Delta D_c(t_k)$$

Equation 19

$$D^{len}_{a,c,m,k} = \rho_{len}(\Delta t_k) D^{len}_{a,c,m,k-1} + \omega_{len} I_{a,c,m,k} \Delta D_c(t_k)$$

Equation 20

where:

- $D^{rap}_{a,c,m,k}$ and $D^{len}_{a,c,m,k}$ are the fast and slow load components at the end of interval *k*.
- $\rho_{rap}(\Delta t_k)$ and $\rho_{len}(\Delta t_k)$ are the retention factors for the two components during the interval. Their values range from 0 to 1.
- ω_{rap} and ω_{len} are the proportions of the new dose assigned to each component and satisfy $\omega_{rap}+\omega_{len} = 1$.

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- $I_{a,c,m,k}$ equals 1 when *m* is the active muscle group during the interval and 0 for the remaining groups.
- $\Delta D_c(t_k)$ is the elementary temporal dose generated during the interval.

The cumulative load of the muscle group at the end of the interval is the sum of its two components:

$$D_{a,c,m,k} = D^{rap}_{a,c,m,k} + D^{len}_{a,c,m,k}$$

Equation 21

At the beginning of the activity, both components equal zero. In each interval, recovery is applied to all muscle groups that retain cumulative load, including those that are not active at that time. The new dose is then distributed between the fast and slow components of the active muscle group.

Accumulation and recovery can therefore occur simultaneously. When the new dose is greater than the reduction produced by recovery, the net load increases. When the dose is small or zero and recovery predominates, the load decreases. The cumulative load does not therefore have to increase continuously.

If the exertion changes direction and requires the action of an antagonist group, the new dose is assigned to that group. The previously active group stops receiving dose but retains its own state and continues to recover. In this way, the time histories of the different functional muscle groups remain separate.

Cumulative load can be consulted at the end of the activity or at any intermediate time. Its time course identifies periods in which exposure increases most rapidly, periods in which appreciable recovery occurs, and the channels or muscle groups that concentrate the greatest temporal demand.

If recovery were not considered, the factors ρ would equal 1 and the procedure would reduce to a sum of the elementary doses. The specific form and parameters of the recovery factors are described in the next section.

10.3.7 Partial recovery between exertions

Temporal integration of the dose makes it possible to estimate how cumulative load evolves while a functional muscle group remains under exertion. In a real activity, however, demand is seldom constant. There may be intervals in which it decreases, changes direction, shifts to an antagonist muscle group, or temporarily ceases. During these periods, the previously accumulated load may be partially reduced through recovery.

Ergoniza Intelligent applies recovery separately to each joint, biomechanical channel, and functional muscle group. In each time interval, the loads previously stored by all muscle groups are attenuated. If a new dose is generated during the same interval, it is then added exclusively to the active muscle group. Recovery is therefore applied continuously, including while a new exposure is being generated: net load increases when the new dose exceeds the reduction produced by recovery and decreases when recovery predominates.

Recovery is represented by two exponential components: a fast component, which loses relevance more rapidly, and a slow component, which retains part of the previous exposure for longer. For an interval of duration Δt, the retention factors for the two components are:

$$\rho_{rap}(\Delta t) = 2^{-\Delta t/h_{rap}}$$

$$\rho_{len}(\Delta t) = 2^{-\Delta t/h_{len}}$$

Equation 22

where h_{fast} and h_{slow} are the half-lives of the fast and slow components, respectively. Half-life is the time required for a component to retain half the value it had at the beginning of the interval.

In the absence of a new dose, the two load components are updated as follows:

$$D^{rap}_{a,c,m}(t+\Delta t) = D^{rap}_{a,c,m}(t) \cdot 2^{-\Delta t/h_{rap}}$$

$$D^{len}_{a,c,m}(t+\Delta t) = D^{len}_{a,c,m}(t) \cdot 2^{-\Delta t/h_{len}}$$

Equation 23

and the remaining load is:

$$D_{a,c,m}(t+\Delta t) = D^{rap}_{a,c,m}(t+\Delta t) + D^{len}_{a,c,m}(t+\Delta t)$$

Equation 24

When a new temporal dose is generated, it is distributed between the two components according to their relative weights:

$$\omega_{rap} + \omega_{len} = 1$$

Equation 25

The complete update, simultaneously including recovery and the new dose, is defined in the preceding section.

For an isolated dose initially distributed between the two components, the proportion remaining after an interval Δt can be expressed as:

$$R(\Delta t) = \omega_{rap} \cdot 2^{-\Delta t/h_{rap}} + \omega_{len} \cdot 2^{-\Delta t/h_{len}}$$

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Equation 26

This expression is useful for representing the recovery of an isolated dose. When the load results from multiple doses generated at different times, the system retains and updates the fast and slow components separately because each may represent a different proportion of the total value at that time.

Using two components with different half-lives is a simplified approximation of a recovery process that does not occur at a single rate. Yates et al. [33] observed that recovery of dynamic muscular endurance progressed rapidly at first, reached approximately 50% after 2 minutes and 15 seconds, was slightly below 90% after 20 minutes, and was best fitted by a three-component exponential curve. Iridiastadi and Nussbaum [34] highlighted the difficulty of representing recovery in intermittent tasks and its dependence on loading conditions. Looft et al. [35] adapted a compartment model to intermittent tasks, and Looft and Frey Law [36] showed that increasing the recovery rate during rest intervals improves the representation of these tasks. Taken together, these findings support distinguishing a faster initial recovery from a more persistent recovery component, without attempting to reproduce the complete physiology of recovery for an individual.

The following values are used in the default configuration of Ergoniza Intelligent:

$$\omega_{rap} = 0.70, h_{rap} = 120\ s$$
$$\omega_{len} = 0.30, h_{len} = 600\ s$$

Thus, 70% of each new dose is assigned to the fast component, which has a half-life of 2 minutes, and the remaining 30% is assigned to the slow component, which has a half-life of 10 minutes. After 120 seconds, the fast component retains 50% of the value it had at the beginning of the interval; after another 120 seconds, it retains 25%, and so on. The slow component evolves in the same way, but with a longer half-life.

Considering both components together, an isolated dose retains approximately 61% of its value after 120 seconds and 17% after 600 seconds. A short break therefore partially reduces the influence of previous exposure, while a longer break produces a greater reduction without eliminating it instantaneously.

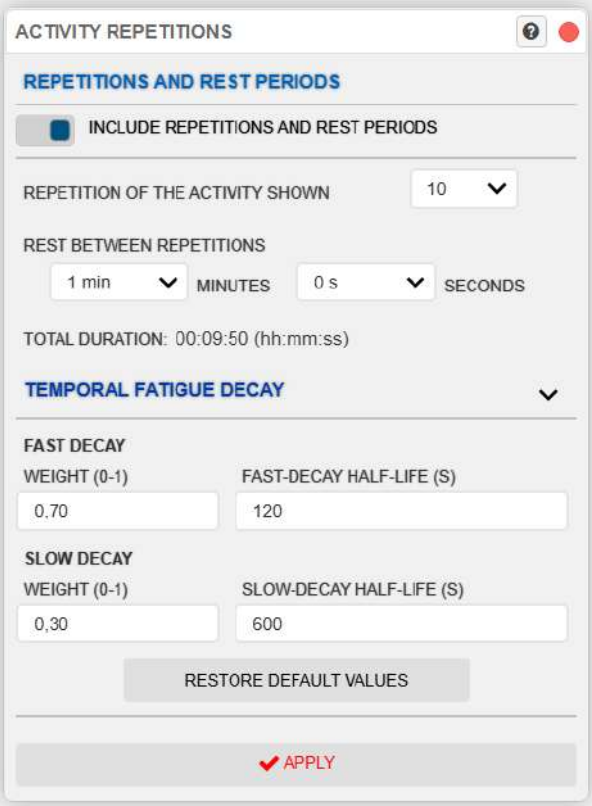


Figure 46: Partial recovery between exertions.

These values should be interpreted as default model parameters, not as universal constants. They were selected to represent an intermediate, prudent behavior: appreciable recovery during the first few minutes without immediately eliminating the effect of repeated exposures. The user can modify them in the **Activity Repetitions** dialog (Figure 46), under **Temporal Fatigue Decay**. The weights of the two components are complementary: when one is changed, the program adjusts the other so that they sum to 1. Half-lives are entered in seconds. In *Multiple Postures* mode, this dialog can be opened from the **Timing** button on the left sidebar or from the clock icon at the bottom of the **Exertion** panel shown in Figure 38. The **Temporal Fatigue Decay** section is collapsed by default; click its heading to expand it.

This procedure distinguishes between activities that generate similar doses but have different time distributions. A sequence of repeated exertions with short breaks will normally produce a greater cumulative load than the same sequence separated

[33] J. W. Yates et al., "Recovery of Dynamic Muscular Endurance," *European Journal of Applied Physiology and Occupational Physiology* 56 (1987): 662–67, doi:10.1007/BF00424807.
[34] Iridiastadi and Nussbaum, "Muscle Fatigue and Endurance during Repetitive Intermittent Static Efforts: Development of Prediction Models."
[35] Looft, Herkert, and Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks."
[36] John M. Looft and Laura A. Frey-Law, "Adapting a Fatigue Model for Shoulder Flexion Fatigue: Enhancing Recovery Rate during Intermittent Rest Intervals," *Journal of Biomechanics*, 2020, doi:10.1016/j.jbiomech.2020.109762.

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by longer rest periods. Similarly, alternation between antagonist muscle groups allows each group to recover partially during the intervals in which the opposing group acts.

The calculated partial recovery should not be interpreted as complete restoration of muscle capacity or as an individual physiological simulation. Its purpose is to represent, in an ergonomic and operational manner, the progressive loss of relevance of previous exposure as demand on a biomechanical channel decreases. This approach is especially important in intermittent tasks, in which the time history of exertion, rest periods, and distribution of active periods determine the final cumulative load [37].

10.3.8 Consideration of repetitive tasks and rest between repetitions

Many work activities are not performed as a single isolated sequence, but as work cycles that repeat several times. In these cases, cumulative exertion load depends not only on what occurs within a cycle, but also on the repetitions performed previously and the rest between them. Two activities with the same base cycle may produce different results if one is repeated several times without sufficient breaks while the other includes longer recovery periods.

When analyzing repetitive activities, Ergoniza Intelligent interprets the entered posture sequence as a representative cycle of the activity. Within each repetition, the time course of the temporal dose and cumulative load is calculated for each joint, biomechanical channel, and functional muscle group using the procedure described in the preceding sections.

The user selects the activity repetition to be represented and analyzed in the dialog shown in Figure 46. If repetition ***N*** is selected, the program assumes that ***N-1*** identical cycles have been completed before the displayed sequence, with the configured rest between them. The selected repetition therefore begins with the residual load from previous repetitions.

Conceptually, let $D_{a,c,m}^{rap,inicio(i)}$ represent the fast component of cumulative load at the beginning of repetition i, $D_{a,c,m}^{rap,fin(i)}$ represent the fast component of cumulative load at the end of repetition i, $D_{a,c,m}^{len,inicio(i)}$ represent the slow component of cumulative load at the beginning of repetition i and $D_{a,c,m}^{len,fin(i)}$ represent the slow component of cumulative load at the end of repetition i. The first repetition begins with a cumulative load of zero:

$$D_{a,c,m}^{rap,inicio(1)} = D_{a,c,m}^{len,inicio(1)} = 0$$

Subsequent repetitions, however, may begin with a residual load. If T_d is the duration of the rest between cycles, the state at the beginning of repetition ***i+1*** is obtained by applying recovery to the state at the end of repetition ***i***:

$$D_{a,c,m}^{rap,inicio(i+1)} = D_{a,c,m}^{rap,fin(i)} \cdot 2^{-T_d/h_{rap}}$$

$$D_{a,c,m}^{len,inicio(i+1)} = D_{a,c,m}^{len,fin(i)} \cdot 2^{-T_d/h_{len}}$$

Equation 27

The total load at the beginning of the new repetition is:

$$D_{a,c,m}^{inicio(i+1)} = D_{a,c,m}^{rap,inicio(i+1)} + D_{a,c,m}^{len,inicio(i+1)}$$

Equation 28

If the rest is short, a substantial portion of the load accumulated at the end of the cycle is therefore retained at the beginning of the next cycle. If the rest is longer, the residual load is smaller. Recovery is applied separately to each functional muscle group, so some groups may begin the new repetition with a greater residual load than others.

During each repetition, load evolves as recovery is applied continuously and the temporal doses generated by the exertions in the sequence are added. Consequently, a later repetition may show higher values than the first even though the biomechanical cycle is identical, because part of the exposure produced in previous cycles remains at the beginning of the repetition.

This approach distinguishes three common situations. An activity repeated with short rest periods will tend to produce progressive accumulation, especially in muscle groups engaged repeatedly. With intermediate breaks, load will increase during exertion periods and partially decrease during rest periods. With longer breaks, residual load may decrease considerably before the next cycle, reducing the cumulative effect of previous repetitions.

[37] Sonne and Potvin, "A Modified Version of the Three-Compartment Model to Predict Fatigue during Submaximal Tasks with Complex Force-Time Histories."

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10.3.8.1 Duration of the represented activity

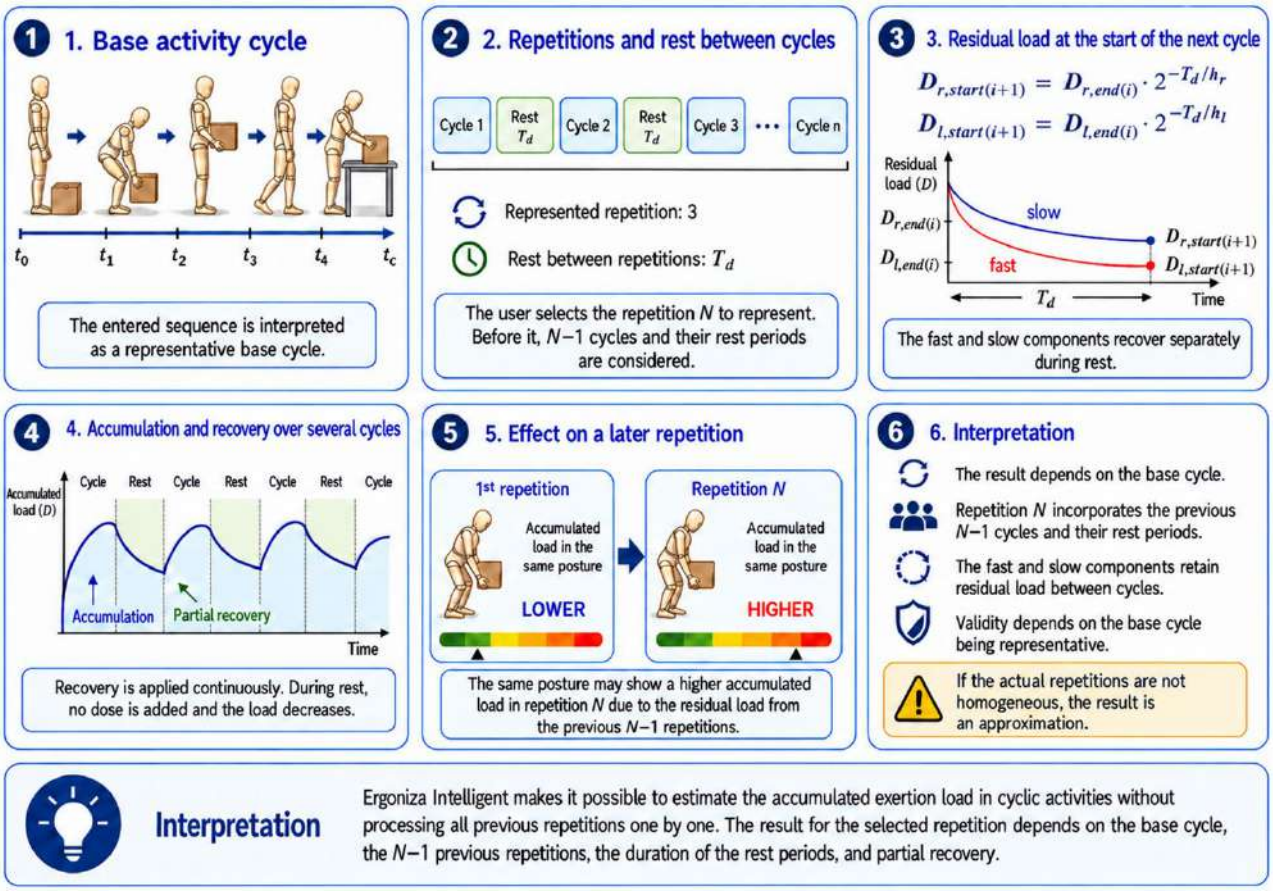
If each cycle has a duration T_c, the rest between cycles is T_d and repetition ***N*** is selected in the dialog shown in Figure 46, the time elapsed from the beginning of the first repetition to the end of the displayed repetition is:

$$T_{actividad} = N \cdot T_c + (N-1)T_d$$

Equation 29

A total of **N-1** rest periods are considered because no rest is added after the represented repetition. The configuration dialog displays this total duration so the user can verify the period of activity being represented.

Considering repetitions and rest periods is especially relevant in intermittent tasks. In these tasks, instantaneous exertion intensity alone does not describe the actual demand of the activity. The time distribution of exertions, duration of active periods, repetition frequency, and duration of breaks determine the final cumulative load. This approach is consistent with fatigue-recovery models applied to tasks with complex force-time histories and with studies showing that recovery during rest periods modifies the development of fatigue in submaximal and intermittent exertions[38].



(Generated using AI)

Figure 47: Consideration of repetitive tasks and rest between repetitions.

The result for a given repetition should be interpreted in the context of the complete work-rest cycle. The same posture may have a low cumulative load during the first repetition and a higher value when it occurs after several cycles with insufficient recovery (Figure 47). The analysis therefore assesses both the demand produced within the base cycle and the effect of repeating it over time.

[38] Xia and Frey Law, "A Theoretical Approach for Modeling Peripheral Muscle Fatigue and Recovery"; Ma et al., "A New Simple Dynamic Muscle Fatigue Model and Its Validation"; Sonne and Potvin, "A Modified Version of the Three-Compartment Model to Predict Fatigue during Submaximal Tasks with Complex Force-Time Histories"; Looft, Herkert, and Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks."

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The **Include Repetitions and Rest Periods** option shown in Figure 46 activates the cumulative exertion load calculation. When the option is selected and repetition 1 is chosen, the program analyzes the first execution of the sequence, without previous repetitions or rest between cycles, but accounting for accumulation and recovery within the sequence itself. When a later repetition is selected, the previous cycles and rest periods are also included.

If the option is cleared, Ergoniza Intelligent does not calculate or display cumulative exertion load. Instantaneous joint exertion and the other biomechanical results continue to be calculated normally, but the time history of the activity is not projected through this indicator.

10.3.9 Final result and ergonomic interpretation

The result of the calculation is a cumulative exertion load for each joint, biomechanical channel, and functional muscle group. This dimensionless quantity represents the state of cumulative exposure produced by the temporal doses generated during the activity, after accounting for the continuous partial recovery described in the preceding sections.

When cumulative load is expressed as a percentage, a value of 100% corresponds to the preventive threshold defined by the model. Values above 100% indicate that this threshold has been exceeded under the analyzed conditions. The threshold does not represent an exact physiological boundary or necessarily imply that fatigue or injury has occurred; rather, it identifies a situation requiring particular attention during the ergonomic assessment.

When the **Calculate Max. Recommended Exertion (%)** control in the **Epidemiological Priority Adjustment** section of the **Biomechanical Analysis Settings** dialog is enabled, Ergoniza Intelligent multiplies the base cumulative load by the weighting factor for the body region. The displayed percentage and its comparison with the 100% threshold therefore incorporate the selected epidemiological priority. This adjustment is described in Section 10.1.1.1.1.

Load is kept separate for each joint, biomechanical channel, and functional muscle group. To obtain the summary value for a biomechanical channel, Ergoniza Intelligent selects the greatest cumulative load among the muscle groups associated with its directions of action. Similarly, the overall value displayed for a joint corresponds to the channel with the greatest cumulative load. Loads from antagonist muscle groups or different biomechanical channels are therefore not added together.

The result should be interpreted in disaggregated form. A high value at a joint identifies the anatomical region subject to the greatest exposure, while the biomechanical channel and functional muscle group identify more precisely the type of exertion that generated it. For example, a high cumulative load in shoulder abduction does not have the same interpretation as a high load in flexion, extension, or rotation. Likewise, a high load in the wrist extensors represents a different exposure from that generated by the flexors or the groups associated with radial and ulnar deviation.

The result should not be confused with instantaneous relative exertion. A channel may show high exertion at a particular time yet maintain a moderate cumulative load if the exposure is brief or sufficient recovery periods are available. Conversely, moderate instantaneous exertions may produce a high cumulative load when sustained for a prolonged period or repeated frequently without sufficient recovery.

Cumulative exertion load therefore distinguishes situations that cannot be adequately differentiated by analyzing a single posture or by using the maximum exertion value reached. Two activities may have the same maximum instantaneous exertion in a given channel yet produce different cumulative loads if they differ in exposure duration, number of repetitions, time distribution of exertions, or rest duration. This characteristic is particularly relevant in cyclic or intermittent tasks.

From an ergonomic perspective, the results identify the joints, biomechanical channels, and functional muscle groups that concentrate the greatest temporal exposure. This information can be used to compare design alternatives, modify load distribution, reduce the duration of particular phases, introduce rest periods, redistribute exertions between the two sides of the body, or prevent the same muscle group from being engaged repeatedly without sufficient recovery.

When several channels have high values, the complete exposure pattern should be analyzed. The load may be concentrated in a single body region or distributed among several joints and muscle groups. In the first case, preventive measures should focus primarily on reducing demand in the affected channel. In the second, a broader review of task organization, movement sequence, work height, load location, or repetition frequency may be necessary.

Figure 48 shows a specific posture within a repeated activity. The active instantaneous indicator appears at the top of the **Exertion** panel. Its name depends on the controls selected in the **Biomechanical Analysis Settings** dialog:

- The **% Maximum Recommended Exertion** is displayed when **Calculate Max. Recommended Exertion (%)** is enabled.

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- The **% Recommended Joint Capacity** is displayed when the preceding control is disabled and **Include Cumulative Exertion Load in Recommended Exertion (%)** remains enabled.
- The **% Maximum Isometric Capacity Used** is displayed when both controls are disabled.

The value corresponds to the represented time.

When the 3CC-r model is involved, the recommended moment used as a reference may reflect the previous exertion and recovery history of the muscle group. However, this instantaneous indicator differs from the **cumulative exertion load** displayed at the bottom of the panel and calculated using the temporal-dose procedure described in this section.



Figure 48: Interpretation of cumulative exertion load.

Cumulative exertion load is displayed at the bottom of the panel. In the example, repetition 10 has been selected and a one-minute rest has been set between repetitions. The result displayed during that repetition therefore incorporates the effect of the base cycle, the nine previous repetitions, and the partial recovery occurring during the rest periods. The same posture may consequently have different cumulative loads depending on whether it appears in the first repetition or after several work cycles.

Consequently, cumulative exertion load should be interpreted as a relative, preventive, and comparative indicator of temporal biomechanical exposure. It is not a direct measurement of individual physiological fatigue or a probability of injury. Its value lies in showing how intensity, duration, repetition, and partial recovery interact, and in helping the ergonomist identify where and how exposure can be reduced.

10.4 The 3CC-r model and Recommended Moment in multiple-posture analysis

The three-compartment model, known as **3CC** (*Three-Compartment Controller*), was proposed as a simplified representation of the development of peripheral muscle fatigue and recovery during an activity[39]. Specific parameters for different joint regions were subsequently obtained[40]. The **3CC-r** model is a modification designed to represent intermittent tasks more accurately by using a multiplier that increases recovery during rest periods[41]. This modification has also been studied specifically for the shoulder joint[42].

The model assumes that, at each instant, muscle capacity is distributed among three states: a **resting** fraction available for activation, an **active** fraction used to generate the required moment, and a **fatigued** fraction that is temporarily unavailable. A controller regulates transitions between the resting and active states to meet demand. During exertion, part of the active capacity moves to the fatigued state at a fatigue rate specific to the body region. When exertion decreases or ceases, fatigued capacity progressively recovers and returns to the resting state. In 3CC-r, this recovery is increased during intervals without demand by recovery multiplier ***r***.

Ergoniza Intelligent applies the model separately to each joint, biomechanical channel, and functional muscle group. At each instant, it calculates relative demand by comparing the magnitude of the required moment with the individualized maximum isometric capacity corresponding to the posture, worker characteristics, and active muscle group. Antagonist muscle groups, such as flexors and extensors, maintain independent states because the activation or recovery of one should not be confused with that of the opposing group.

To use 3CC-r as a recommended criterion rather than a direct prediction of exhaustion, Ergoniza Intelligent incorporates a preventive margin. The time course used to obtain the recommended moment sets the preventive limit at 50% of the maximum endurance time estimated by the model itself for the body region. The result is therefore not intended to predict the

[39] Xia and Frey Law, "A Theoretical Approach for Modeling Peripheral Muscle Fatigue and Recovery."
[40] Frey-Law, Looft, and Heitsman, "A Three-Compartment Muscle Fatigue Model Accurately Predicts Joint-Specific Maximum Endurance Times for Sustained Isometric Tasks."
[41] Looft, Herkert, and Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks."
[42] John M. Looft and Laura A. Frey Law, "Adapting a Fatigue Model for Shoulder Flexion Fatigue: Enhancing Recovery Rate during Intermittent Rest Intervals," *Journal of Biomechanics*, 2020, doi:10.1016/j.jbiomech.2020.109762.

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exact time at which a particular worker will reach fatigue, but to establish a preventive reference that accounts for the time history of exertion and recovery.

$$M_{rec,3CC-r,z}(t) = M_{iso,z}(t) \cdot f_{3CC-r,z}(t)$$

Equation 30

where:

- $M_{rec,3CC-r,z}$ (t) is the **3CC-r Recommended Moment** for joint or channel **z** at time **t**;
- $M_{iso,z}$(t) is the moment corresponding to the individualized maximum isometric capacity for joint or channel *z* at time *t*; and
- $f_{3CC-r,z(t)}$ is the recommended capacity fraction obtained from the model, with values between 0 and 1.

As the fatigued fraction increases, this factor decreases and the recommended moment is reduced. During periods of lower demand or rest, recovery may cause the factor to increase again. If **Include Cumulative Exertion Load in Recommended Exertion (%)** is disabled or the 3CC-r calculation is unavailable, the factor takes a value of 1 and the reference moment equals the individualized maximum isometric capacity.

When the activity contains several repetitions, the 3CC-r states are propagated from one cycle to the next and recovery during the rest periods between them is considered. This procedure is independent of the **cumulative exertion load** calculation described in Section 10.3: the 3CC-r state is not obtained from the cumulative dose calculated using the MET curves, although both procedures use the same biomechanical sequence.

Where applicable, the resulting reference moment is combined with the epidemiological-priority adjustment described in Section 10.1.1.1.1. The displayed indicator depends on the controls selected in the **Biomechanical Analysis Settings** dialog. When **Calculate Max. Recommended Exertion (%)** is enabled, the program displays **% Maximum Recommended Exertion**. When this control is disabled but **Include Cumulative Exertion Load in Recommended Exertion (%)** is enabled and a valid 3CC-r result is available, it displays **% Recommended Joint Capacity**. If both controls are disabled, or a 3CC-r calculation is unavailable, it displays **% Maximum Isometric Capacity Used**.

10.5 Conceptual differences between single- and multiple-posture analysis

Single-posture analysis, described in Section 10.2, and the temporal procedures used in multiple-posture analysis, described in Sections 10.3 and 10.4, represent the activity in different ways and provide complementary results.

In *single-posture* analysis, the activity is represented by a fixed postural configuration and specified loads. First, Ergoniza Intelligent calculates the **maximum isometric capacity** of each joint and biomechanical channel, individualized according to the worker's characteristics. The exertion duration and repetitiveness conditions then determine the recommended **Maximum Voluntary Contraction Percentage** (%MVC). Applying this percentage to the individualized maximum isometric capacity yields the **Base Maximum Moment** used to assess the posture.

The result can be displayed as **% Recommended Joint Capacity** or, when **Calculate Max. Recommended Exertion (%)** is enabled in the **Epidemiological Priority Adjustment** section of the **Biomechanical Analysis Settings** dialog, as **% Maximum Recommended Exertion**. The recommended joint capacity percentage relates the required moment to the Base Maximum Moment. The maximum recommended exertion percentage then weights this result according to the prevalence of musculoskeletal disorders assigned to the body region. In both cases, the indicator assesses the analyzed posture under the temporal conditions declared by the user, but does not reconstruct the time course of exertion throughout an activity.

In *multiple-posture* analysis, Ergoniza Intelligent obtains, on the one hand, an exertion indicator for the represented time. When **Include Cumulative Exertion Load in Recommended Exertion (%)** is enabled in the **Biomechanical Analysis Settings** dialog, the required moment is compared with the 3CC-r Recommended Moment, which may vary according to the exertion and recovery history of the active muscle group. If this option is disabled or the 3CC-r calculation is unavailable, the required moment is compared with the individualized maximum isometric capacity. The operation of the 3CC-r model is explained in Section 10.4.

The indicator name depends on the controls selected in the **Biomechanical Analysis Settings** dialog. **% Maximum Recommended Exertion** is displayed when **Calculate Max. Recommended Exertion (%)** is enabled. If this control is disabled, **% Recommended Joint Capacity** is displayed when **Include Cumulative Exertion Load in Recommended Exertion (%)** is enabled and a valid 3CC-r result is available. If the latter option is also disabled or the 3CC-r calculation is

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unavailable, **% Maximum Isometric Capacity Used** is displayed. Although the indicator corresponds to the represented time, when 3CC-r is involved it may reflect the previous history of exertion and recovery.

Cumulative exertion load, on the other hand, is calculated using the independent procedure described in Section 10.3. In each interval, the required moment is related to the individualized maximum isometric capacity, transformed into a temporal dose using Maximum Endurance Time curves, and integrated while accounting for partial recovery. The cumulative dose does not modify the state calculated by 3CC-r. When **Calculate Max. Recommended Exertion (%)** is enabled, the displayed cumulative-load value incorporates the weighting corresponding to the prevalence of musculoskeletal disorders assigned to the body region.

The active instantaneous indicator and cumulative exertion load should therefore not be interpreted as equivalent quantities. The former expresses the relationship between the moment required in the visible posture and the applicable reference moment under the selected settings. When 3CC-r is involved, that reference moment may reflect the previous history of exertion and recovery. Cumulative load, by contrast, represents the state of biomechanical exposure reached up to the time being considered as a result of the complete time course of exertions and partial recovery.

When several cycles of a repetitive activity are considered, the cumulative load for the selected repetition also incorporates residual load from previous repetitions and recovery during rest periods between cycles.

A posture may have a high value for the active instantaneous indicator yet produce a moderate cumulative load if it is held briefly or followed by sufficient recovery. Conversely, moderate instantaneous exertions may produce a high cumulative load when sustained for prolonged periods or frequently repeated using the same muscle group without sufficient recovery.

Single-posture analysis is appropriate when the activity can reasonably be represented by a characteristic posture and simple temporal conditions. Multiple-posture analysis is more appropriate when there are postural changes, load variations, alternation between muscle groups, phases of lower demand, rest periods, or repeated cycles.

The two procedures should be used as complementary tools. Instantaneous indicators identify the postures or joints closest to their reference limits, while cumulative load identifies the biomechanical channels and muscle groups that concentrate the greatest exposure over the course of the activity.

10.6 Displaying exertions on the model

Ergoniza Intelligent can use colors on the human model to represent exertion-analysis results for the different joints and body regions. This representation helps quickly locate the regions subject to the greatest biomechanical demand, although its purpose is primarily indicative and it should be interpreted together with the numerical values shown in the result charts and tables.

The display is activated with the **Visualization on the Model** button, identified by an eye-shaped icon at the bottom of the **Exertion** panel (Figure 49). When activated, the joints represented on the model change color according to the level reached by the selected quantity.

The model's original color identifies joints whose values are below the **Warning** level. As a value approaches or reaches that level, the color transitions toward orange. When the **Risk** level is reached or exceeded, the joint is shown in red. The gradual color transition also indicates proximity to these levels.

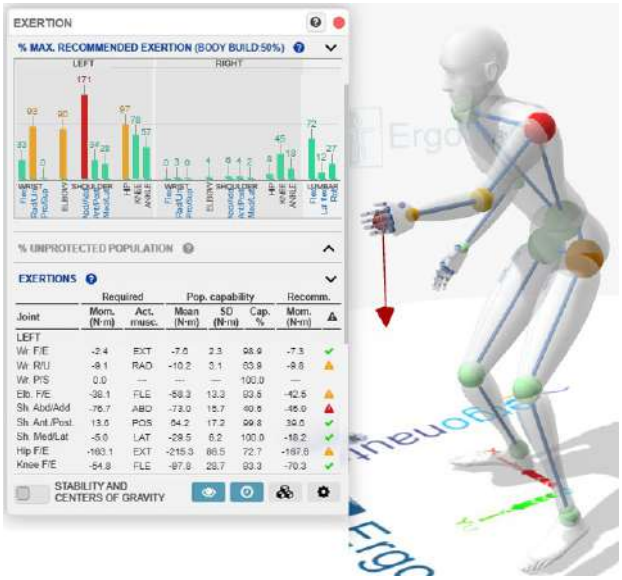


Figure 49: Visualization of exertions on the model.

The **Warning** and **Risk** values depend on the represented indicator. For % Recommended Joint Capacity, % Maximum Recommended Exertion, % Maximum Isometric Capacity Used, and cumulative exertion load, the initial values are 85% for Warning and 100% for Risk. These thresholds can be customized in the **Visualization** section of the **Settings** panel. When % Unprotected Population is represented, the action and danger limits established for that procedure are applied; these depend on the sex of the analyzed population and the configuration defined by the user.

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10.6.1 Display in single-posture analysis

In single-posture analysis, the colors represent the indicator active in the Exertion panel:

- **% Recommended Joint Capacity**, when **Calculate Max. Recommended Exertion (%)** is disabled.
- **% Maximum Recommended Exertion**, when **Calculate Max. Recommended Exertion (%)** is enabled.
- **% Unprotected Population**, when this assessment procedure is selected.

The representation corresponds exclusively to the posture and loads displayed at that time. It therefore locates the joints with the greatest instantaneous demand. In this case, it does not represent cumulative exposure over the course of an activity.

10.6.2 Display in multiple-posture analysis

In multiple-posture analysis, the display can show either the active instantaneous indicator or cumulative exertion load.

The instantaneous indicator is calculated for the posture corresponding to the current position of the animation cursor. Depending on the assessment procedure and selected settings, it may correspond to **% Recommended Joint Capacity**, **% Maximum Recommended Exertion**, **% Maximum Isometric Capacity Used**, or **% Unprotected Population**. As the sequence is played through, the colors show how instantaneous biomechanical demand varies from one posture to another.

The **cumulative exertion load** represents the state of exposure reached up to the displayed time. Its colors therefore depend not only on the visible posture, but also on previous exertions, their duration and repetition, partial recovery, and, where applicable, previous repetitions of the analyzed cycle.

To assign a color to a joint in the cumulative-load display, Ergoniza Intelligent uses the greatest value among its associated biomechanical channels. The color therefore locates the most unfavorable channel for each joint without adding exposures corresponding to different movements or muscle groups.

When visualization is activated in multiple-posture mode and cumulative-load calculation is enabled, the program asks the user to choose which of the two quantities to represent: the active instantaneous indicator or cumulative exertion load. If cumulative-load calculation is not enabled, the instantaneous indicator is displayed directly.

Figure 49 shows an example of this color coding. The image quickly identifies the most prominent joints, but the colors alone do not indicate which biomechanical channel or functional muscle group produced the result. The corresponding tables and charts must be consulted for that interpretation.

The final assessment should not be based exclusively on the representation on the model. The numerical values, active quantity, time within the sequence, duration and repetition of exertions, partial recovery, and actual operating conditions of the activity should be considered together.

10.7 Limitations of joint exertion as a sole risk indicator

Joint-exertion analysis estimates the biomechanical demand required to maintain a posture, balance external loads, or perform an activity represented by a posture sequence. However, its results should not be interpreted as a complete ergonomic risk assessment. Joint exertion is a relevant dimension of the analysis, but it does not account for every mechanism through which a posture or task may adversely affect the musculoskeletal system.

Joint moments, % Recommended Joint Capacity, % Maximum Recommended Exertion, % Maximum Isometric Capacity Used, % Unprotected Population, and cumulative exertion load provide different perspectives on biomechanical demand. None of these indicators alone expresses the probability of injury or a musculoskeletal disorder.

A task may also be unfavorable because of extreme or sustained postures, loading of passive structures, localized tissue compression, surface contact, instability, repetitive low-amplitude movements, vibration, environmental conditions, work organization, or the individual's characteristics.

Exertion-analysis results should therefore be interpreted together with the other available ergonomic indicators, technical observation of the activity, and the actual conditions under which the work is performed.

10.7.1 Limitations in single-posture analysis

Calculating joint moments and reaction forces estimates the biomechanical demand required to maintain a posture and balance the applied loads. However, a low joint moment does not necessarily mean that the posture is acceptable or that the joint is free from mechanical loading.

Figure 50 shows an example of this difference. In posture **A**, load ***P*** and the weight of the body segments have appreciable moment arms relative to the wrist, elbow, and shoulder. The corresponding muscle groups must generate internal

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moments to maintain equilibrium, so the analysis clearly identifies the resulting biomechanical demand.

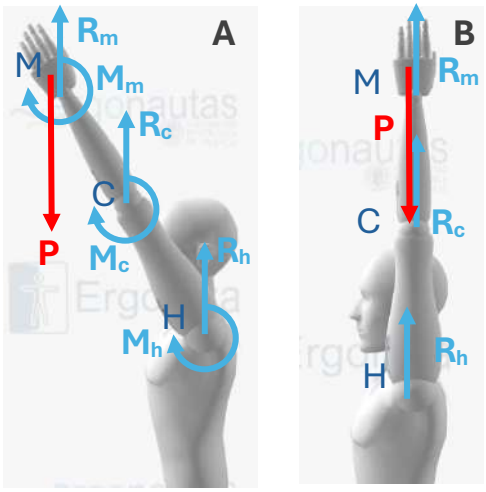


Figure 50: Effect of aligning a load with body segments on the calculated joint moments.

In posture **B**, the line of action of the external load is approximately aligned with the arm segments. Its moment arm relative to the joints is consequently small, and the joint moments produced by that load may be low. However, the load may still transmit compressive or tensile forces through the joints, and the posture may require muscular stabilization. In addition, the weight of the body segments themselves may continue to generate moments if their centers of mass are not aligned with the joints.

Keeping the arm raised above the head may also be unfavorable because of deviation from a comfortable functional posture, reduced range-of-motion margin, compression or tension in certain tissues, and the demand created by maintaining the posture for a prolonged period. These effects may be relevant even when the calculated joint moments are low.

Similarly, a joint may be close to its functional limits even when the required moment is low. In these situations, risk does not necessarily result from a high contraction of the primary muscle groups, but from the postural configuration, loading of passive structures, or reduction in the available range-of-motion margin.

The indicators used by Ergoniza Intelligent also have specific limitations. **% Recommended Joint Capacity** compares the required moment with the **Base Maximum Moment**, customized according to the worker's characteristics and the temporal conditions declared for the action, but does not account for every musculoskeletal risk mechanism. **% Maximum Recommended Exertion** incorporates weighting based on the prevalence of musculoskeletal disorders by body region, but should be interpreted as a preventive-priority index rather than an individual estimate of susceptibility or injury probability. **% Unprotected Population** is based on population capacity distributions and does not replace assessment of the specific worker's characteristics.

For these reasons, posture assessment should not be based exclusively on joint-exertion limits. Ergoniza Intelligent supplements force and moment analysis with postural-risk assessment and other evaluation procedures. This combination can detect unfavorable situations even when the posture geometry produces low joint moments.

10.7.2 Limitations in multiple-posture analysis, the 3CC-r model, and cumulative exertion load

The 3CC-r Recommended Moment, described in Section 10.4, is obtained through mathematical modeling of the development of exertion and recovery during an activity. Its parameters are derived from experimental results, and its application assumes that the posture sequence, its durations, and the rest periods adequately represent the analyzed task. Variation in the 3CC-r Recommended Moment should therefore be interpreted as a preventive estimate of the available recommended capacity, not as a direct measurement of physiological state, actual strength loss, or residual fatigue in a particular worker.

In activities represented by multiple postures, cumulative exertion load estimates how relative exertion intensity, duration, repetition, and partial recovery combine over time. This information helps identify the joints, biomechanical channels, and functional muscle groups that concentrate the greatest temporal exposure. However, the result should be interpreted as a relative ergonomic indicator of biomechanical exposure, not as a direct measurement of an individual's muscle fatigue.

The actual physiological response may differ from the estimates provided by both temporal procedures. It depends on factors that these procedures cannot fully represent, such as training, physical condition, preexisting conditions, neuromuscular variability, task experience, temperature, required precision, motivation, prior pain, or cumulative exposure during the rest of the workday. Two people exposed to the same sequence may respond differently, and the tolerance of the same person may vary at different times or on different workdays.

Cumulative load also does not represent a direct probability of injury. A high value indicates that a particular biomechanical channel concentrates substantial temporal exposure relative to the adopted preventive criterion, but cannot by itself predict

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the development of a musculoskeletal disorder. The relationship between exposure and harm depends on the interaction of biomechanical, individual, environmental, and organizational factors.

The result depends on the validity of the input data and preceding biomechanical calculations. If the postures, external loads, supports, durations, or equilibrium conditions do not adequately represent the actual activity, both the 3CC-r Recommended Moment and the resulting cumulative load will be affected by those simplifications.

In posture sequences, Ergoniza Intelligent can calculate linear and angular velocities and accelerations from the model's time course. However, these kinematic quantities are not incorporated fully and systematically as inertial forces and moments in the joint-exertion calculation. The biomechanical exertion model should therefore be considered primarily static or quasi-static. It should be applied with particular caution to tasks involving rapid movements, impacts, relevant accelerations, or abrupt changes in direction because the dynamic loads produced in these situations may not be fully represented.

The transformation of relative exertion into a temporal dose is based on population Maximum Endurance Time curves and a preventive criterion applied to them. These relationships provide a useful ergonomic approximation but do not exactly reproduce the endurance of every worker or every possible contraction condition.

The partial recovery incorporated into the calculation is also an approximation. The model represents the progressive loss of cumulative load using two temporal recovery components. Actual physiological recovery may differ according to body region, muscle group, intensity and duration of previous exposure, alternation between tasks, and the individual's characteristics.

When an activity is projected over several repetitions, the calculation assumes that the base cycle and defined rest periods are representative. If the actual repetitions vary substantially in posture, loads, duration, or rest periods, the result should be interpreted as an approximation of the actual activity.

The procedure focuses on temporal exposure associated with joint moments. It does not incorporate every musculoskeletal risk mechanism, such as localized tissue compression, mechanical contact, vibration, grip quality, pressure on joint structures, fine repetitive movements of the hands and fingers, cold exposure, psychosocial factors, or overall work organization.

Consequently, both the instantaneous indicator, when it incorporates the 3CC-r Recommended Moment, and cumulative exertion load should be used as preventive and comparative tools. The former assesses how the recommended moment changes over the sequence according to the estimated history of exertion and recovery. The latter shows how biomechanical exposure is distributed and accumulated, identifies the channels and muscle groups with the greatest exposure, and assesses the potential effect of changes in posture, loads, work sequence, repetitions, or rest periods. The final interpretation should consider these indicators together with observation of the actual task, characteristics of the worker population, and the other relevant ergonomic factors.

11 POSTURAL RISK ANALYSIS

Maintaining or repeating awkward or sustained postures during an activity can cause fatigue and contribute to the development of musculoskeletal disorders. Adopting unsuitable postures is therefore one of the risk factors that should be considered when evaluating an activity. Reducing exposure to these postures is an essential preventive measure.

Ergoniza Intelligent evaluates postures using three complementary procedures based on different criteria. The first calculates the discomfort associated with the posture of each joint using the procedure described in Section 11.2. The other two apply the REBA and LUBA methods to assess the risk arising from adopted postures; both are described in Section 11.3.

11.1 Joint ranges of motion

Assessing the discomfort associated with a posture requires knowledge of joint ranges of motion. Joint range of motion (ROM) is the angular arc through which a joint can move between its limiting positions under defined measurement conditions. Its extent depends on the geometry of the articular surfaces, the elasticity of periarticular tissues (such as the capsule, ligaments, and muscles), and, when measured actively, the neuromuscular capacity to initiate and control the movement.

ROM values show considerable interindividual variability because they depend on multiple intrinsic and extrinsic factors:

- **Age**, because of the progressive loss of elasticity and structural changes in cartilage and soft tissues.
- **Sex**, because morphological and hormonal differences affect ligament laxity and muscle strength.

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- **Previous conditions or surgery**, which may limit movement through shortening, adhesions, or loss of joint congruence.
- **Physical-activity and training level**, because regular flexibility or strength exercise can increase or stabilize ROM.

ROM values also depend on how they are measured. The following are distinguished according to the measurement procedure:

- **Passive ROM**: the evaluator moves the joint without active participation by the subject. This method explores the maximum mobility allowed by the soft tissues and joint capsule and is particularly sensitive to deep restrictions (e.g., capsulitis or fibrosis).
- **Active ROM**: the subject generates the movement through voluntary muscle contraction. Its extent is usually smaller than passive ROM because strength, neuromuscular control, and coordination are also involved.

The difference between the two depends on the joint, movement direction, posture, and measurement procedure.

Some anatomical regions also function as joint complexes consisting of several joints that cooperate to produce a global movement. For example, when the arm is raised above the horizontal plane, shoulder flexion or abduction results from the coordinated action of the glenohumeral, sternoclavicular, and acromioclavicular joints together with scapulothoracic movement. Measurement may attempt to isolate a specific joint (for example, the glenohumeral joint) or record the global movement of the complex, allowing its components to participate synergistically. The two procedures measure different phenomena and may provide different ROM values.

The biomechanical and clinical literature includes numerous studies and manuals intended to quantify joint ranges of motion in healthy adults. However, the published values vary considerably because of differences in measurement protocols, characteristics of the populations studied, and definitions of reference points and movement planes, among other factors. Posture during measurement also influences the results. Many studies determine ROM with the person lying down or in standardized clinical positions, while measurements performed while standing during an activity are less common.

Following a review of the literature[43] and consideration of both the characteristics of the Ergoniza Intelligent biomechanical model and the postural-discomfort measurement procedure described in Section 11.2.1, the ROM values used are based primarily on data from Kee and Karwowski (2001, 2004). Modifications supported by other literature sources have been introduced for some upper- and lower-limb movements. This results in some differences from the ROM values defined in standards such as ISO 11226.

Finally, these ranges of motion are used as reference values for calculating postural discomfort, but do not by themselves limit the postures that the Ergoniza Intelligent biomechanical model can adopt.

[43] G. S. Fleisig et al., "Active Range of Motion of the Shoulder: A Cross-Sectional Study of 6635 Subjects," *JSES International* 7, no. 1 (2023): 132.137, doi:10.1016/j.jseint.2022.09.008; AAOS, *Joint Motion: Method of Measuring and Recording* (Chicago, 1965); L. Cocchiarella and GBJ Andersson, *Guides to the Evaluation of Permanent Impairment. 5th Ed.*, *The Journal of Bone and Joint Surgery-American Volume*, 5ª (Chicago, 2001), doi:10.2106/00004623-200109000-00050; D. Norkin, C. & White, C. Norkin, and D. White, *Measurement of Joint Motion: A Guide to Goniometry*, *Physiotherapy Canada*, 5ª (Philadelphia: F. A. Davis Company, 2016); D. C. Boone and S. P. Azen, "Normal Range of Motion of Joints in Male Subjects," *Journal of Bone and Joint Surgery - Series A* 61, no. 5 (1979): 756–59, doi:10.2106/00004623-197961050-00017; B. L. Greene and S. L. Wolf, "Upper Extremity Joint Movement: Comparison of Two Measurement Devices," *Archives of Physical Medicine and Rehabilitation* 70, no. 4 (1989): 288–90; Izge Gunal et al., "Normal Range of Motion of the Joints of the Upper Extremity in Male Subjects, with Special Reference to Side," *Journal of Bone and Joint Surgery* 78, no. 9 (1996): 1401–4, doi:10.2106/00004623-199609000-00017; WB Greene and JD Heckma, eds., *The Clinical Measurement of Joint Motion* (Rosemont, IL: American Academy of Orthopaedic Surgeons, n.d.); Jaiyoung Ryu et al., "Functional Ranges of Motion of the Wrist Joint," *Journal of Hand Surgery* 16, no. 3 (1991): 409–19, doi:10.1016/0363-5023(91)90006-W; S. Svenningsen et al., "Hip Motion Related to Age and Sex," *Acta Orthopaedica Scan* 60, no. 1 (1989): 97–100, doi:10.3109/17453678909150103; K. E. Roach and T. P. Miles, "Normal Hip and Knee Active Range of Motion: The Relationship to Age," *Physical Therapy* 71, no. 9 (1991): 656–65, doi:10.1093/ptj/71.9.656; C. A. Lantz, J. Chen, and D. Buch, "Clinical Validity and Stability of Active and Passive Cervical Range of Motion with Regard to Total and Unilateral Uniplanar Motion," *Spine* 124, no. 11 (1999): 1082–89, doi:10.1097/00007632-199906010-00007; J. W. Youdas, J. R. Carey, and T. R. Garrett, "Reliability of Measurements of Cervical Spine Range of Motion - Comparison of Three Methods," *Physical Therapy* 71, no. 2 (1991): 98–104, doi:10.1093/ptj/71.2.98; M. Tousignant et al., "Criterion Validity Study of the Cervical Range of Motion (CROM) Device for Rotational Range of Motion on Healthy Adults," *Journal of Orthopaedic and Sports Physical Therapy* 36, no. 4 (2006): 242–48, doi:10.2519/jospt.2006.36.4.242.

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11.2 Postural discomfort

In ergonomic practice, standards and methods such as ISO 11226, EN 1005-4, RULA, and REBA[44] provide criteria for assessing the acceptability of particular postures and prioritizing preventive intervention. They classify angular deviations into intervals and, depending on the method, combine them with factors such as posture duration, muscle use, applied force or load, and coupling quality. The result is usually expressed through discrete risk categories or scores. These procedures are not designed to provide a continuous assessment for every degree of angular variation, nor do they consider every possible joint degree of freedom.

By contrast, **postural discomfort** refers to the **subjective** (perceptual) sensation of discomfort associated with a joint posture, especially when it is maintained under static conditions. It is interpreted as a psychophysical outcome: when joint configurations move away from the neutral or preferred zone, people tend to report less comfort even if the instantaneous mechanical demand (e.g., joint moment) is low.

11.2.1 Measuring postural discomfort: psychophysical basis

Postural discomfort and postural risk are not equivalent quantities. The former estimates a perceptual response associated with joint configuration; risk methods, by contrast, combine different exposure factors to obtain a preventive classification or score.

Ergoniza Intelligent calculates joint forces and moments using the procedure described in Section 8 and assesses biomechanical exertion using the indicators in Section 10. Discomfort analysis supplements those results by quantifying a perceptual component of posture that is not described by joint moments or exertion percentage alone. Interpreting both types of information together provides a more complete characterization of the analyzed situation.

Ergoniza Intelligent expresses postural discomfort on a percentage scale based on regression equations obtained for different joints and movements[45]. These equations come from psychophysical scaling experiments in which participants maintained static postures and rated their comfort using the free-modulus magnitude-estimation method. The responses were then normalized and fitted using regressions that relate joint angle to perceived comfort. In the original studies, most fits had R^2 values above 0.90.

The validity of psychophysical measurements for comfort assessment has been supported by previous studies. This approach integrates participants' subjective perceptions with quantitative analysis methods, providing reliable and representative data. Values obtained through psychophysical scaling show a strong correlation with physiological and biomechanical measures. In addition, statistical normalization of the responses minimizes interindividual variability, making the results applicable to broader populations[46]. Nevertheless, the equations represent the responses of the populations and experimental conditions studied; their results should therefore be interpreted as ergonomic estimates rather than exact predictions for a particular person.

	Men	Women
Extremely Good	100.0	100.0
Very Good	94.1	90.3
Good	86.4	80.7
Moderate	75.8	71.0
Marginal	66.4	58.0
Somewhat Poor	52.1	41.9
Poor	37.4	23.4
Very Poor	20.5	10.8
Extremely Poor	0.0	0.0

Table 5: Geometric-mean normalized comfort scores corresponding to the verbal categories.

To interpret the values obtained from the regression equations, comfort intervals were established based on analysis of the normalized comfort values reported during the experiments. These intervals were defined by converting subjective ratings into qualitative categories and assigning them normalized values on a scale from 0 to 100. The main categories are shown in Table 5.

These values relate the estimate obtained from the comfort equations to a verbal category and facilitate its interpretation. The model distinguishes between men and women, showing significant sex-related differences in the perception of postural comfort. These differences are reflected both in the absolute values in Table 5 and in the distribution of comfort levels across the range of motion. Therefore, Ergoniza Intelligent differentiates its calculation by sex, underscoring the

[44] N. J. Delleman and J. Dul, "International Standards on Working Postures and Movements ISO 11226 and EN 1005-4," *Ergonomics*, 2007, doi:10.1080/00140130701674430.

[45] D. Kee and W. Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures," *Ergonomics* 15, no. 4 (2004): 427–45, doi:10.1080/00140130310001638180; D . Kee and W. Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures," *Ergonomics* 15, no. 44 (2001): 614–48, doi:10.1080/00140130110038044.

[46] J. J. Zwislocki and D. A. Goodman, "Absolute Scaling of Sensory Magnitudes: A Validation," *Perception & Psychophysics* 27, no. 1 (1980): 28–38, doi:10.3758/BF03204312.

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importance of adapting ergonomic analysis to the specific characteristics of each population.

The regression equations and comfort intervals were obtained from postures maintained for 60 seconds. In other words, they measure comfort in relatively prolonged static postures without supporting a load.

To incorporate posture-holding time into discomfort, Ergoniza Intelligent first calculates postural discomfort using a base time of 60 seconds and the regression equations (referred to as **absolute discomfort**). Once absolute discomfort has been obtained, it is modified to account for posture-holding time using the procedures described in Sections 11.2.3 and 11.2.4. These adaptations extend the operational use of the original equations and should be interpreted subject to the limitations described later.

11.2.2 Absolute discomfort assessment

Absolute discomfort is the value estimated for a posture maintained for **60 seconds** without handling external loads. Ergoniza Intelligent calculates it for each joint, movement, and direction using the regression equations described in Section 11.2.1.

The regression equations provide a continuous assessment of the effect of angle on comfort, avoiding artificial changes when discrete thresholds are crossed. They also associate the verbal categories (e.g., very good, good, moderate, marginal, somewhat poor, ...) with a numerical value on the same normalized scale. This correspondence makes it possible to use these categories as interpretable comfort/discomfort references and, above all, to anchor the indicator at a selected level according to interpretation needs and convergence with other frameworks (such as ISO 11226, EN 1005-4, RULA, or REBA).

Ergoniza Intelligent generates an absolute-discomfort indicator as a percentage, with 0% representing the optimal condition and 100% the value that should not be exceeded. This is performed as follows. A regression equation is available for each joint/movement **j**[47]:

$$y(ang) = f_j(ang)$$

Equation 31

where:

- **ang** is the movement angle (in degrees, within the allowable range or ROM),
- **y** is the continuous normalized comfort value predicted by the model.

A comfortable reference value is established so that the indicator starts at 0 in the neutral movement posture (ang=0):

$$y_{ref,j} = y(0) = f_j(0)$$

Equation 32

Finally, a value $\mathbf{C_{k,j}}$ is selected for each joint or body region j as the operational limit, that is, the comfort value associated with 100% discomfort. This value is not the same for all joints; it is established specifically for each body region, movement, movement direction, and analyzed postural condition. Absolute discomfort ($\mathbf{D_j}$) is thus defined as:

$$D_j(ang) = 100\ \frac{y_{ref,j} - y_j(ang)}{y_{ref,j} - C_{k,j}}$$

Equation 33

where $y_{ref,j}$ is the comfort value predicted by the equation for joint j in the neutral posture, $y_j(ang)$ is the comfort value predicted for the analyzed angle, and $C_{k,j}$ is the comfort value used as the reference for the maximum recommended limit at that joint and movement.

As defined, discomfort D_j satisfies the following:

- $D_j(0) = 0$; that is, discomfort at joint j in the neutral posture is 0.
- $D_j(ang) = 100$ when $y_j(ang) = C_{k,j}$; that is, discomfort reaches 100% when the comfort predicted by the regression reaches the value established as the recommended limit.

The value $\mathbf{C_{k,j}}$ is selected in Ergoniza Intelligent using two criteria. First, the recommendations of established standards and methods such as ISO 11226, EN 1005-4, RULA, or REBA are used when they provide relevant recommendations. When the standards do not specify a clear limit or do not consider a particular movement at a joint, the value corresponding to the Marginal verbal category in Table 5 is used.

This approach simultaneously satisfies two requirements: it preserves the progression (slope/curvature) of the regression equations, and therefore of the psychophysical data, while

[47] Kee and Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures"; Kee and Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures."

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allowing an operational threshold (100%) to be established in alignment with an external criterion.

This adjustment does not alter the shape of the regression curve; it only rescales the axis so that the 100 threshold has a meaning aligned with the literature and preventive practice while preserving continuity, angular sensitivity, and differentiation by sex.

Figure 51 shows, as an example, the calculation of absolute discomfort (purple line) in the forearm from the elbow-flexion angle for women. Discomfort is assessed as a percentage and increases as the angle increases. The figure also shows the regression equation (y(x)) and the comfortable reference value (97.77) corresponding to y(0).

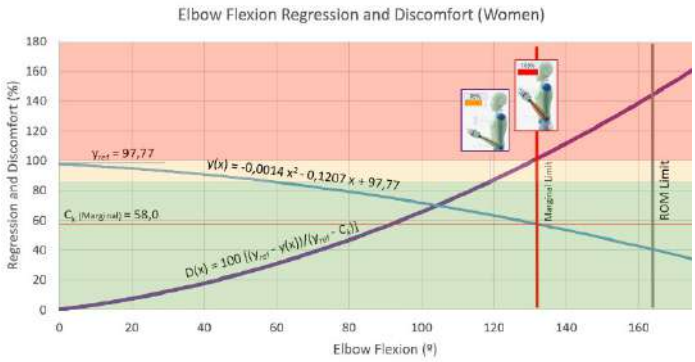


Figure 51: Calculation of postural discomfort in the forearm.

The vertical red line marks the angle at which regression equation y(x) reaches the "Marginal" value (58.0 in Table 5), beyond which comfort begins to deteriorate. In Ergoniza Intelligent, this angle corresponds to 100% discomfort, that is, the point beyond which the posture can be considered uncomfortable. Postures that reach a discomfort value of 85% are shown in orange.

As shown in Figure 51, the discomfort graph extends beyond the joint's maximum ROM. If the user does not limit joint ranges while manipulating the model (by disabling Biological Constraints in the interface), unnatural joint angles adopted by the model may produce inconsistent displayed discomfort values.

11.2.2.1 Considering upper-limb orientation relative to gravity

When assessing upper-limb postural discomfort, Ergoniza Intelligent can consider segment orientation relative to gravity. This option affects the calculation of upper-arm and forearm discomfort and prevents a posture from being assessed solely from the relative angles between segments when trunk inclination substantially changes the actual demand of supporting the limb.

The basic postural-discomfort calculation uses the position of each joint relative to the preceding segment. For the upper arm, this means assessing its position relative to the trunk; for the forearm, its position relative to the upper arm. This approach is suitable in many situations, especially when the trunk is upright. In that posture, an upper arm aligned with the trunk approximately corresponds to an arm hanging vertically toward the floor and is therefore favorably oriented relative to gravity.

However, when the trunk tilts, flexes, or bends laterally, the anatomical and gravitational references no longer coincide. For example, if the trunk is flexed forward, an upper arm parallel to the trunk may be close to the neutral position relative to the shoulder but oriented horizontally in space. In that situation, even though the relative upper-arm-to-trunk angle is small, maintaining the arm requires greater muscle activity to support its weight against gravity. Conversely, an arm forming a large angle relative to the trunk may hang vertically toward the floor and have a more favorable orientation.

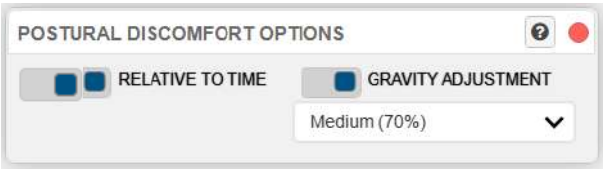


Figure 52: Postural-discomfort calculation options

For this reason, when **Gravity Adjustment** (Figure 52) is enabled, Ergoniza Intelligent introduces a correction to upper-limb postural discomfort. This correction does not replace the joint calculation based on relative angles; it partially reduces the result when the orientation of the upper arm or forearm is clearly favored by gravity.

For the upper arm, the system calculates trunk orientation relative to vertical and upper-arm orientation relative to downward vertical. The influence of the correction increases as the trunk moves away from vertical. When the trunk is nearly upright, the correction is zero or very small because the calculation based on the relative upper-arm-to-trunk angle is sufficiently representative. As the trunk inclines, the actual orientation of the upper arm relative to gravity becomes more important. Its influence is greatest when the arm is aligned with downward vertical and disappears as the arm moves away from that direction.

For the forearm, the system calculates the orientation of the segment between the elbow and wrist relative to downward vertical. The correction is applied only when the forearm is close to a natural hanging orientation toward the floor. As with the upper arm, maximum gravity assistance is considered to

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occur when the forearm is aligned with downward vertical and progressively disappears as it moves away from that direction. This correction is applied less strongly than for the upper arm because elbow discomfort depends primarily on joint flexion, and the mechanical demand associated with forearm weight is also reflected in the biomechanical exertion analysis.

The selector shown in Figure 52 defines the maximum intensity of this correction (to access this dialog, click the gear-shaped button at the bottom of the **Postural Load** panel shown in Figure 59). This value does not directly indicate the reduction in discomfort, but the maximum weight that favorable orientation relative to gravity can have in the final calculation. For example, Medium (70%) indicates that, when the posture meets the conditions for the correction to act at maximum intensity, orientation relative to gravity can appreciably reduce the final assessment, although the relative-joint criterion still contributes to the result.

The correction is applied progressively. For the upper arm, if the trunk is only slightly inclined, the effect of gravity will be small even if the selector has a high value. As trunk inclination increases, the correction approaches the selected maximum weight. For the forearm, the correction is more moderate and is applied according to its orientation relative to downward vertical.

This option does not turn the postural-discomfort calculation into a biomechanical analysis of moments or muscle exertion. Its purpose is to adjust continuous upper-limb postural assessment to account for the fact that the same relative position between segments may involve different demands depending on its orientation relative to gravity. Detailed analysis of moments, joint forces, and maximum recommended exertion percentage is performed in the Biomechanical Exertion module (Section 10).

11.2.3 Time-relative postural discomfort in single-posture analysis

As explained in Section 11.2.2, Ergoniza Intelligent defines **absolute discomfort** as the discomfort measured for a posture held statically for 60 seconds. **Time-relative postural discomfort** is absolute discomfort adjusted to the actual time for which the analyzed posture is held. To calculate relative discomfort, Ergoniza Intelligent uses two different procedures depending on whether a single posture or an activity composed of multiple postures represented in an animation is being analyzed (see Section 11.2.4).

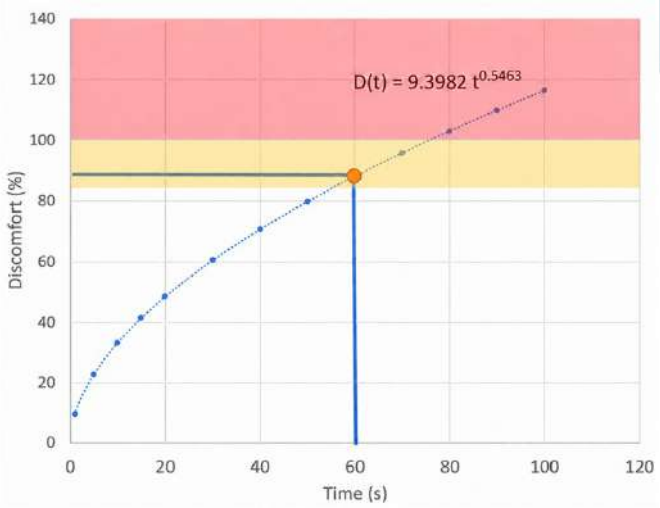


Figure 53: Postural discomfort in the shoulder over time.

When Ergoniza Intelligent is used in **Single Posture** mode, the analysis focuses on the single posture adopted by the biomechanical model. In this case, the user must specify how long the posture is held. To do so, the dialog shown in Figure 42 can be used to specify the duration and repetitiveness of the posture. This dialog can be opened with the **Timing** button on the left sidebar or with the **clock icon** at the bottom of the **Postural Risk** panel shown in Figure 59**.** The following sections explain how relative discomfort is calculated according to the duration and type of posture holding.

The **Postural Discomfort Options** dialog shown in Figure 52 includes an option for excluding posture-holding time from the postural discomfort calculation. When **Relative to Time** is disabled, the discomfort results shown by Ergoniza Intelligent correspond to absolute discomfort, that is, the discomfort calculated for postures held for 60 seconds.

11.2.3.1 Postural discomfort in a static posture

Static or sustained postures are those in which the body maintains the same posture for a more or less prolonged period, with little or no movement. Unlike dynamic or repetitive postures, which involve continuous movement or alternating body segments, static postures involve sustained muscle contraction that can cause localized fatigue.

In Ergoniza Intelligent, the user can use the dialog shown in Figure 42 to specify how long the posture is held by selecting "*Does not repeat; it is sporadic but is sustained for a certain duration*" in the Repetitiveness box. Once absolute discomfort has been calculated as explained in 11.2.2, its variation according to the duration of a static posture is considered to obtain time-relative discomfort. For this purpose, Ergoniza

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Intelligent uses an adaptation of the Law and Avin[48] models (Equation 34) to predict Maximum Endurance Time (or "endurance time") in static muscle contractions according to relative exertion intensity (%MVC) for different joints. Each equation follows a power law whose parameters, obtained by regression from experimental data, characterize how rapidly tolerance decreases when a posture is maintained at each joint.

$$t = b_0\, MVC^{b1}$$

Equation 34

The scale constant (b_0) establishes the temporal reference point, while the exponent (b_1) defines the decay curve: the more negative b_1 is, the more sharply endurance falls as relative load increases. Specific values for the ankle, knee, elbow, shoulder, and torso, together with a general case, allow these models to finely adapt the estimate of static fatigue to the anatomical and functional variability of each joint. This provides a robust tool for assessing risk and designing exposure guidelines that is suitable for sustained postures because it reproduces the nonlinear relationship observed between sustained exposure and exertion tolerance.

Ergonza Intelligent adapts the models to postural discomfort over time as follows. First, it solves for MVC, yielding Equation 35.

$$D(t, b_0, b_1) = MVC(t, b_0, b_1) = \left(\frac{t}{b_0}\right)^{\frac{1}{b1}}$$

Equation 35

This curve is anchored to the discomfort value measured after the posture has been held for 60 seconds (absolute discomfort), yielding Equation 36.

$$D_{est}(t, b_1) = D_{absoluto}\left(\frac{t}{60}\right)^{\frac{1}{b1}}$$

Equation 36

Anchoring the curve to the value measured at 60 seconds ensures that the model reflects the experimentally verified discomfort level (absolute discomfort), while also capturing the change in fatigue as posture-holding time varies. The resulting value is again expressed as a percentage; 100% discomfort corresponds to the level at and above which the posture can be considered uncomfortable.

Figure 53 shows an example of the type of potential curve obtained with this approach. The curve was calculated for the shoulder joint when absolute discomfort in the adopted posture is 88%. The power exponent, fitted using joint-specific muscle endurance data, causes discomfort to increase rapidly during the first few seconds and more slowly as the posture continues. This behavior is consistent with *endurance time* [49]48F and with subjective perception curves for static postures, in which the sensation increases more slowly after a rapid initial "jump" in fatigue.

11.2.3.2 Postural discomfort in a repetitive posture

To account for posture duration when assessing discomfort in repetitive or intermittent tasks—those in which a static posture is held for only part of a work cycle and alternates with pauses—while in **single-posture analysis mode**, Ergoniza Intelligent uses a method based on the Threshold Limit Values (TLVs) for muscle loading from the ACGIH[50] (American Conference of Governmental Industrial Hygienists).

When the Ergoniza Intelligent user selects *Occurs periodically in repeating movement cycles* in the **Repetitiveness** box shown in Figure 42, the duration of the work cycle, the duration of one exertion, and the number of exertions performed in each work cycle must be entered. Using these data, Ergoniza Intelligent calculates the effective time for which the posture is held within the cycle (t), that is, the number of seconds per minute (or other units of time) during which the posture—or postures similar to the one analyzed—is held. For example, if a person performing a cyclic task holds a certain posture for 10 seconds in every 60-second cycle, the *duty cycle* (DC) is 16.7%.

The procedure described in Section 11.2.3.1 is used to calculate the discomfort associated with the effective posture-holding time. This discomfort value is then corrected using the TLV factor corresponding to the observed *duty cycle*, as shown in Equation 37. In this equation, j represents the joint being analyzed, *t* is the effective posture-holding time, and DC is the percentage of effective posture-holding time relative to the total cycle time.

[48] Law and Avin, "Endurance Time Is Joint-Specific: A Modelling and Meta-Analysis Investigation."
[49] Rohmert, "Ermittlung von Erholungspausen F¨ur Statische Arbeit Des Menschen"; C J De Luca, "Myoelectrical Manifestations of Localized Muscular Fatigue in Humans.," *Critical Reviews in Biomedical Engineering* 11, no. 4 (1984): 251–79; Jack Ogutu and Woojin Park, "The Relationship between Perceived Discomfort of Static Posture Holding and Posture Holding Time," *Work* 52, no. 1 (2015): 19–30, doi:10.3233/WOR-141946.
[50] ACGIH, "Ergonomic Threshold Limit Value (TLV) for Hand Activity Level (HAL) and Muscle Fatigue."

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The parameter α_j is the time exponent associated with joint *j*. Its function is to modulate the influence of posture-holding time on estimated discomfort. This parameter is obtained from the exponent $b_{1,j}$ of the Frey-Law and Avin Maximum Endurance Time models, expressed in Equation 34 as $t = b_{0,j} \cdot MVC\hat{}b_{1,j}$. In these models, $b_{1,j}$ is negative and specific to joint *j*. Solving for relative intensity and normalizing the result against the 60-second reference time used by Ergoniza Intelligent to calculate absolute discomfort yields the positive exponent $\alpha_j = -1 / b_{1,j}$. This exponent allows absolute discomfort, measured for 60 seconds of posture holding, to be rescaled to the actual exposure time t.

$$D_{rep,j}(t, DC, \alpha_j) = \frac{D_{absoluto,j}\left(\frac{t}{60}\right)^{\alpha_j}}{\left[-0.143 \cdot ln\left(\frac{DC}{100}\right) + 0.066\right]}$$

Equation 37

The resulting value is again expressed as a percentage; 100% discomfort corresponds to the level at and above which the posture can be considered uncomfortable.

When the assessed joint belongs to the torso or lower extremities, Ergoniza Intelligent applies the same internal adaptation coefficients described for calculating recommended %MVC in repetitive tasks (see Table 3 in Section 10.2.3). These coefficients are not region-specific TLVs, but rather a conservative approximation based on relative differences in fatigue resistance described in the literature.

11.2.3.3 Postural discomfort in a momentary or unsustained posture

The procedures described in the preceding sections make it possible to assess postural discomfort when a posture is held for a certain time or forms part of a repetitive task. In these cases, exposure time is a fundamental variable because holding a posture away from neutral can cause fatigue, reduce exertion tolerance, and progressively increase the perception of discomfort.

However, not all postures last long enough to be interpreted as sustained postures. In some tasks, the posture being analyzed may be a momentary, transient, or nearly instantaneous position. Ergoniza Intelligent considers this to be the case when the user selects **Does not repeat; it is sporadic and is sustained only momentarily** in the dialog shown in Figure 42.

In these situations, directly applying a temporal fatigue model is inappropriate. A posture adopted for an instant does not consume endurance time in a way comparable to a posture held for several seconds or minutes. However, this does not mean that every momentary posture should be considered acceptable. A joint positioned near the limits of its range of motion can cause discomfort, tissue strain, reduced functional margin, or exposure to an inadvisable postural configuration, even if that position is not held for long.

For this reason, Ergoniza Intelligent uses a specific procedure for momentary or unsustained postures. Instead of estimating the increase in discomfort caused by posture-holding time, it assesses the instantaneous proximity of each joint to the end of its range of motion. The objective is to identify postures that, although brief, place joints in regions far from the comfortable functional zone.

The complete mathematical formulation of the procedure and the rationale for the parameters used are provided in Appendix 15.2.

11.2.3.3.1 Biomechanical and psychophysical basis

The procedure is based on the principle that a joint's range of motion does not constitute a homogeneous comfort zone. A joint may be within its anatomical range yet still adopt an uncomfortable position or one close to a functional limit. The literature therefore distinguishes between the total anatomical range, comfortable movement zones, and resting or maximally comfortable positions.

Kee and Karwowski's studies[51] of joint angles of isocomfort show that the relationship between joint angle and perceived comfort depends on the joint, movement, and direction being assessed. Comfort does not decrease in the same way in all body regions: some show an appreciable reduction with relatively moderate deviations, while others tolerate wider ranges. Consequently, applying a single angular threshold to every joint is inappropriate.

This distinction is also reflected in the concepts of **Comfort Range of Motion** (CROM) and **Range of Rest Posture** (RRP). CROM represents the portion of the range of motion associated with comfortable conditions, while RRP identifies a more restricted zone associated with resting or maximally comfortable postures. Apostolico et al. [52] place RRP within

[51] Kee and Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures"; Kee and Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures."

[52] Alessandro Apostolico et al., "Postural Comfort Evaluation: Experimental Identification of Range of Rest Posture for Human

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CROM for the neck, shoulder, elbow, wrist, and ankle, and emphasize its usefulness for characterizing the most favorable joint configurations.

Other studies [53] have used the body's angular configuration to classify postural discomfort or to develop predictive models using regression and machine learning [54]. These findings support the proposition that joint positions contain information relevant to estimating the discomfort associated with a specific posture, although they do not eliminate individual variability or turn that estimate into a direct physiological measurement.

Finally, discomfort arising from angular configuration must be distinguished from discomfort associated with mechanical demand. Studies of discomfort and joint moment show nonlinear, joint-dependent relationships [55]. Ergoniza Intelligent addresses this second dimension through the exertion analysis described in Section 10. The momentary-posture procedure developed here is therefore limited to instantaneous angular configuration and does not estimate the muscular demand produced by joint moments or external loads.

11.2.3.3.2 General calculation principle

For momentary or unsustained postures, Ergoniza Intelligent calculates an instantaneous discomfort indicator based on the relative position of each joint within its effective range of motion. A central functional zone in which momentary discomfort equals 0 is defined for each joint and degree of freedom. Once the limit of that zone is exceeded, the indicator increases continuously as the joint approaches the end of its range of motion.

This calculation does not estimate fatigue caused by sustaining a posture over time. Its purpose is to detect brief or transient joint configurations that are close to functional limits, even when they are not held for a prolonged period.

For each joint or body region *j* and each degree of freedom or movement *d*, the normalized angular position is first calculated:

$$r_{j,d} = min\left(\frac{|ang_{j,d}|}{ROMef_{,j,d}}, 1\right)$$

Equation 38

where:

- ***ang**$_{j,d}$* is the angle adopted by joint or region *j* in movement and direction *d*, measured relative to the neutral or reference position;
- ***ROMef**$_{j,d}$* is the effective range of motion used as a reference in that direction;
- ***r**$_{j,d}$* is the relative position within the effective range: r=0 represents the neutral posture and r=1 represents the effective ROM limit. If the angle exceeds that limit, the normalized value remains at 1.

The effective ROM does not physically modify the movement permitted by the biomechanical model. It is used only as a reference for measuring proximity to the joint limit and may differ from the base ROM because of age-related adjustment factors. The procedure used to obtain it is detailed in Appendix 15.2.

Next, a threshold ***r0**$_{j,d}$* is defined to represent the upper limit of the central functional zone. While the joint remains below this value, momentary discomfort is considered zero or negligible. When the joint exceeds this threshold, discomfort increases progressively.

$$r_{j,d} \leq r0_{j,d} \rightarrow \mathrm{D}_{\mathrm{puntual,j,d}} = 0$$

$$r_{j,d} > r0_{j,d} \rightarrow \mathrm{D}_{\mathrm{puntual,j,d}} > 0$$

Equation 39

The value ***r0**$_{j,d}$* is not the same for every joint. Ergoniza Intelligent calculates it separately from the relative sensitivity assigned to each joint and degree of freedom. For the upper extremities, neck, and torso, this sensitivity is obtained from the maximum scores and angular intervals in the LUBA method [56]. LUBA is based on assessments of static postures and must not be interpreted as a model specifically developed

Articular Joints," *International Journal on Interactive Design and Manufacturing* 8 (2014): 109–20, doi:10.1007/s12008-013-0186-z.

[53] Jaejin Hwang and Kyung Sun Lee, "Classification of Whole-Body Postural Discomfort Using Cluster Analysis," *International Journal of Environmental Research and Public Health* 17, no. 22 (2020): 8314, doi:10.3390/ijerph17228314.

[54] Jinwon Lee, Jaejin Hwang, and Kyung Sun Lee, "Prediction and Comparison of Postural Discomfort Based on MLP and Quadratic Regression," *Journal of Occupational Health* 63, no. 1 (2021): e12292, doi:10.1002/1348-9585.12292.

[55] Takanori Chihara, Taiki Izumi, and Akihiko Seo, "Perceived Discomfort Functions Based on Joint Moment for Various Joint Motion Directions of the Upper Limb," *Applied Ergonomics*, 2014, doi:10.1016/j.apergo.2013.04.016.

[56] D Kee and W Karwowski, "LUBA: An Assessment Technique for Postural Loading on the Upper Body Based on Joint Motion Discomfort and Maximum Holding Time.," *Applied Ergonomics* 32, no. 4 (August 2001): 357–66, http://www.ncbi.nlm.nih.gov/pubmed/11461037.

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for instantaneous postures. Therefore, Ergoniza Intelligent does not use LUBA to calculate momentary discomfort directly, but only as a source of relative information about the sensitivity of each joint and degree of freedom. For the lower extremities, an alternative procedure based on the Kee and Karwowski isocomfort equations is used[57]. The method for determining $r0_{j,d}$, including the use of LUBA for the upper extremities, neck, and torso and the alternative isocomfort-equation procedure for the lower extremities, is described in Appendix 15.2.

The value $r0_{j,d}$ is determined separately for each joint, movement, direction, and postural condition. For the upper extremities, neck, and torso, Ergoniza Intelligent combines the maximum score and the entry angle for the highest-scoring interval in the LUBA method[58]. LUBA is not applied as a method for assessing the momentary posture or as a temporal model; it is used only to establish relative differences in sensitivity among joints and degrees of freedom. For the lower extremities, an alternative procedure based on the Kee and Karwowski isocomfort equations is used. The detailed calculation of this parameter is provided in Appendix 15.2.

The 100% momentary-discomfort value does not coincide with the absolute ROM limit, but with a zone close to that limit. In the current implementation, 100% is reached when the joint reaches 95% of its effective ROM (r100=0.95).

This criterion introduces a preventive margin: a momentary posture does not need to reach the exact end of the range of motion to be classified as extreme. If the movement continues beyond that threshold and approaches the effective ROM limit even more closely, the indicator can exceed 100%, up to a bounded maximum. In the current implementation, the maximum value considered is 125%. The transition function used to produce this progressive increase in the indicator is detailed in Appendix 15.2.

11.2.3.3.3 Interpreting the result

The momentary discomfort calculated using this procedure is expressed as a percentage. A value close to 0% indicates that the joint is within a central functional zone or sufficiently far from its joint limit. A value close to 100% indicates that the joint is approaching an extreme zone of its effective ROM. Values above 100% indicate that the operational threshold for an extreme posture has been exceeded and that the joint is moving even closer to the effective ROM limit.

The result must be interpreted as an indicator of instantaneous postural deviation, not as an indicator of cumulative fatigue. It therefore does not replace the temporal analysis of sustained postures or the analysis of repetitive tasks. Nor does it replace the biomechanical analysis of joint exertion: a posture may have low angular discomfort but high muscular demand due to an external load, or conversely, it may have a low joint moment but high postural deviation.

The purpose of the procedure is to provide a continuous indicator for identifying and comparing brief or transient postures close to functional limits. Its interpretation must be combined with the analysis of joint exertion, observation of the actual task, and, when appropriate, temporal assessment of sustained or repetitive postures.

11.2.4 Time-relative postural discomfort in multiple postures

When an activity is represented by a posture sequence (**Multiple Postures** mode in the interface), postural discomfort cannot be obtained simply by analyzing each posture in isolation. The same joint configuration may be more or less unfavorable depending on prior exposure: reaching an extreme posture for the first time, holding it for a certain period, and repeatedly returning to similar angular configurations are not equivalent.

For this reason, Ergoniza Intelligent includes a specific procedure for assessing time-relative postural discomfort in posture sequences. The sequence is analyzed as a temporal succession of postures to estimate how sustained or repeated exposure to unfavorable postural configurations modifies the discomfort associated with the posture observed at each instant.

11.2.4.1 Scope and quantities used to calculate discomfort in posture sequences

The calculation of discomfort in posture sequences maintains the conceptual distinction between postural discomfort and biomechanical exertion. Postural discomfort is based on the angular deviation of joints from functional or comfortable zones, whereas biomechanical exertion is calculated from moments, joint forces, and muscle capacity. This distinction is necessary because a posture can be posturally unfavorable without involving a high percentage of maximum voluntary

[57] Kee and Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures"; Kee and Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures."

[58] Kee and Karwowski, "LUBA: An Assessment Technique for Postural Loading on the Upper Body Based on Joint Motion Discomfort and Maximum Holding Time."

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contraction, while high muscle demand can occur in postures that are not particularly extreme from an angular perspective. This distinction is consistent with Kee and Karwowski's[59] studies of discomfort associated with joint deviations and with the findings of Frey Law et al.[60], which show that the relationships among perception, pain, exertion, and maximum holding time are not equivalent in static and dynamic tasks.

For each joint or degree of freedom assessed, the procedure considers three main components:

- the **momentary discomfort** of the current posture at each instant, corresponding to a posture adopted briefly or not sustained;
- the **absolute discomfort** (see Section 11.2.2), which represents the discomfort that the current postural configuration would produce if held for the baseline time used in the psychophysical comfort equations;
- the **equivalent time of prior postural exposure**, which summarizes cumulative exposure to angularly similar postures while accounting for distance from neutrality and partial recovery over time.

In general terms, the result can be expressed as:

$$D_{final,j}(t) = F_j(D_{puntual,j}(t), D_{60,j}(t), T_{eq,j}(t))$$

Equation 40

where:

- $D_{final,j}(t)$ is the final postural discomfort of joint or degree of freedom *j* at time *t*;
- $D_{puntual,j}(t)$ is the discomfort corresponding to the current posture when it is considered momentary or unsustained;
- $D_{60,j}(t)$ is the absolute discomfort of the current posture (for a 60-second exposure);
- $Teq,j(t)$ is the equivalent time of prior exposure relevant to the current posture;
- Fj is the combination function that integrates the baseline effect of the posture and the cumulative temporal effect.

The **equivalent time** $Teq,j(t)$ is not a direct sum of seconds. Only prior exposures that occurred in configurations sufficiently far from neutral, angularly similar to the current posture, and recent enough that their effects have not been fully recovered make a relevant contribution. The system therefore answers the following question: for how much prior time has the joint been in postures similar to the current one and with sufficient postural significance to influence present discomfort?

Once the equivalent time has been obtained, Ergoniza Intelligent applies a nonlinear temporal transformation to the absolute discomfort of the current posture at each instant. The use of a nonlinear relationship is supported by experimental evidence on sustained static postures. Ogutu and Park[61] observed that the development of discomfort over time is not generally described by a linear function and that a power function with an additive term adequately represents most of the individual discomfort-time sequences analyzed.

The general procedure can be summarized as follows: at each instant, the effective posture is calculated; functional joint angles are obtained; momentary discomfort (see Section 11.2.3.3) and absolute discomfort (see 11.2.2) are determined; prior exposure equivalent to similar postural configurations is estimated; and time-relative postural discomfort is finally obtained using a combination that always preserves the baseline value of the current posture, so that the final result cannot be lower than the momentary discomfort of the current posture.

The result must be interpreted as an ergonomic indicator of equivalent postural exposure in posture sequences. It is not a clinical prediction, a direct probability of injury, an exact physiological measure of muscle fatigue, or a quantified reduction in force-generating capacity. Its main purpose is to compare design alternatives, detect moments of unfavorable postural exposure, and assess the effect of sustaining or repeating joint configurations outside the comfortable functional zone.

The following sections describe the calculations used to determine time-relative postural discomfort in multiple-posture sequences.

11.2.4.2 Measuring postural exposure time

To estimate the effect of sustaining or repeating postures in a sequence, Ergoniza Intelligent processes the posture sequence over time and calculates, for each joint and degree of freedom analyzed, how long it remains in different angular zones. The objective is not to store each posture as an

[59] Kee and Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures"; Kee and Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures."

[60] Laura A. Frey Law et al., "Relationships between Maximum Holding Time and Ratings of Pain and Exertion Differ for Static and Dynamic Tasks," *Applied Ergonomics* 42 (2010): 9–15, doi:10.1016/j.apergo.2010.03.007.

[61] Ogutu and Park, "The Relationship between Perceived Discomfort of Static Posture Holding and Posture Holding Time."

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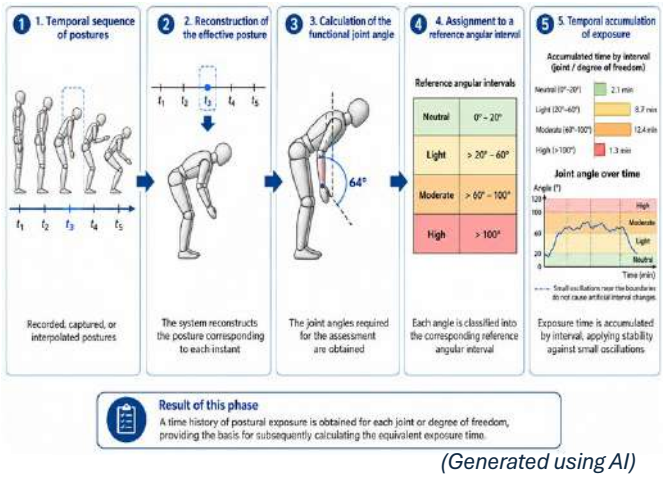
independent case, but to construct a temporal representation of cumulative postural exposure.

At each instant assessed, the system reconstructs the model posture. The functional joint angles required to assess postural discomfort are calculated from that posture.

For each joint and degree of freedom *j*, the resulting angle is assigned to a reference ***angular interval***. These angular intervals are not the same size for every movement; instead, they are adapted to the functional range and postural sensitivity of each joint. Consequently, the same angular difference in degrees does not necessarily produce the same effect in every body region.

The system accumulates exposure time in each angular interval. To prevent artificial changes when an angle fluctuates slightly around the boundary between two intervals, a stability criterion is applied to prevent spurious switching between adjacent intervals. This produces a more stable representation of calculated exposure and prevents small numerical fluctuations from repeatedly changing the assigned interval.

The recorded temporal exposure provides the basis for subsequently calculating the equivalent time $T_{eq,j}(t)$. At this stage, therefore, the system is not summing past discomfort scores; it is measuring how long the joint has remained in angular configurations that may be relevant to the current posture.



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Figure 54: Temporal accumulation of exposure by angular interval and degree of freedom.

This stage produces a temporal history of postural exposure for each joint and degree of freedom (Figure 54). In subsequent steps, this history makes it possible to weight prior exposure according to three criteria: similarity to the current posture, distance from neutrality, and partial recovery over time.

11.2.4.3 Equivalent postural exposure time

Using the temporal history of postural exposure described above, Ergoniza Intelligent calculates, for each joint and degree of freedom *j*, an equivalent exposure time $T_{eq,j}(t)$. This equivalent time represents the portion of prior exposure that remains relevant when interpreting discomfort in the current posture.

Equivalent time does not necessarily coincide with the chronological time during which the joint has been outside the neutral posture. Two exposures of equal duration may have different effects if one occurred in a configuration very similar to the current posture and the other in a different angular configuration. Similarly, a recent exposure has a greater effect than an older one because partial recovery occurs over time.

The calculation of $T_{eq,j}(t)$ is therefore based on three criteria:

- angular similarity between the prior exposure and the current posture;
- the postural relevance of the angular interval, so that positions close to neutral contribute little or nothing;
- the remaining temporal memory, which progressively reduces the weight of older exposures.

In general terms, equivalent time can be expressed as:

$$T_{eq,j}(t) = \sum_{k<t} \Delta t_k \cdot S_j(\theta_k, \theta(t)) \cdot A_j(\theta_k) \cdot M_j(t - t_k)$$

Equation 41

where:

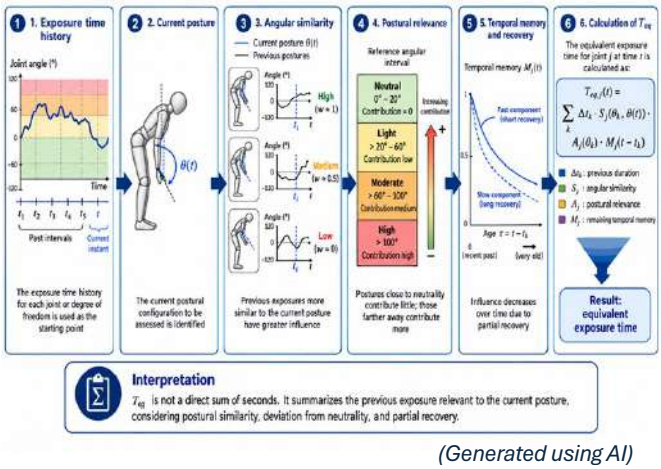
- $T_{eq,j}(t)$ is the equivalent postural exposure time of joint or degree of freedom j at time t;
- Δt_k is the duration of prior time interval k;
- θ_k is the representative joint angle during that prior interval;
- $\theta(t)$ is the current joint angle at the time being assessed;
- $S_j(\theta_k, \theta(t))$ is the angular-similarity function between the prior exposure and the current posture;
- $A_j(\theta_k)$ is the postural activation associated with the prior angle;
- $M_j(t-t_k)$ is the remaining temporal memory after the time elapsed since the prior exposure.

The following sections define the different components of this equation.

This approach avoids treating all prior exposures as equivalent. A prior posture very similar to the current one makes an important contribution to equivalent time if it occurred recently and was sufficiently far from neutral. In contrast, a neutral

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posture, an angularly different posture, or an older exposure makes a reduced or zero contribution (Figure 55).



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Figure 55: Equivalent postural exposure time.

Consequently, $T_{eq,j}(t)$ must be interpreted as a temporal measure weighted by postural similarity, angular relevance, and partial recovery. Its function is to indicate how much prior exposure time can be considered posturally equivalent to having sustained configurations similar to the current one.

11.2.4.4 Angular similarity between prior exposure and the current posture

Not all prior exposure contributes equally to discomfort in the current posture. A prior exposure is relevant only if it occurred in an angular configuration similar to that adopted by the joint at the time being analyzed.

For each joint and degree of freedom *j*, Ergoniza Intelligent compares the current angle *θ(t)* with the representative angles of the previously occupied reference angular intervals. The contribution of each prior exposure is weighted using an angular-similarity function:

$$S_j(\theta_k, \theta(t)) \in [0,1]$$

Equation 42

where $S_j = 1$ represents maximum similarity between the prior exposure and the current posture, and $S_j = 0$ indicates that the prior exposure is not considered relevant to the posture being assessed.

Conceptually, this similarity depends on angular distance:

$$\delta_j = \mid \theta_k - \theta(t) \mid$$

Equation 43

The smaller δ_j, the greater the influence of the prior exposure. For example, a prior exposure at 50° of flexion has a greater effect on a current posture at 55° than on one at 15°.

The comparison is made within the same functional direction of movement. For example, a prior exposure in flexion is not interpreted as equivalent exposure for a current posture in extension.

Angular similarity does not itself incorporate posture severity. Its function is to identify which prior exposures are posturally similar to the current one. Severity is subsequently introduced through absolute discomfort $D_{60,j}(t)$ and through the postural relevance of the corresponding reference angular interval.

11.2.4.5 Postural relevance of the reference angular interval

In addition to similarity to the current posture, Ergoniza Intelligent considers whether the prior exposure occurred in a postural configuration sufficiently far from neutral. Time spent in positions close to the neutral posture should not contribute significantly to cumulative discomfort, even when those positions are repeated or sustained for a certain period.

Each reference angular interval therefore has an associated postural-relevance function:

$$A_j(\theta_k) \in [0,1]$$

Equation 44

where $A_j(\theta_k)$ represents the postural significance of prior angle θ_k for joint and degree of freedom j.

Values close to 0 indicate that the configuration is within or close to the neutral postural zone and therefore contributes little to the equivalent time. Values close to 1 indicate that the configuration is sufficiently far from neutral to be considered relevant postural exposure.

Conceptually:

$$A_j(\theta_k) \approx 0 \quad \text{if } \theta_k \text{ is close to neutral}$$
$$A_j(\theta_k) \to 1 \quad \text{if } \theta_k \text{ moves away from neutral}$$

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11.2.4.6 Temporal memory and partial recovery

Prior postural exposure does not retain the same influence on discomfort in the current posture indefinitely. After leaving an unfavorable configuration, part of its effect must be considered to recover progressively. To represent this behavior, Ergoniza Intelligent applies a temporal-memory function $M_j(\tau)$, where τ is the time elapsed between the prior exposure and the instant being analyzed:

$$\tau = t - t_k$$

Equation 45

The memory function is defined as a combination of two exponential components:

$$M_j(\tau) = w_r \cdot 2^{-\frac{\tau}{T_r}} + w_l \cdot 2^{-\frac{\tau}{T_l}}$$

Equation 46

where:

- $M_j(\tau)$ is the memory factor applied to the prior exposure of degree of freedom *j* after time τ;
- w_r is the weight of the fast component;
- T_r is the half-life of the fast component;
- w_l is the weight of the slow component;
- T_l is the half-life of the slow component.

The half-lives are the times required for the influence of a prior exposure to be reduced by half.

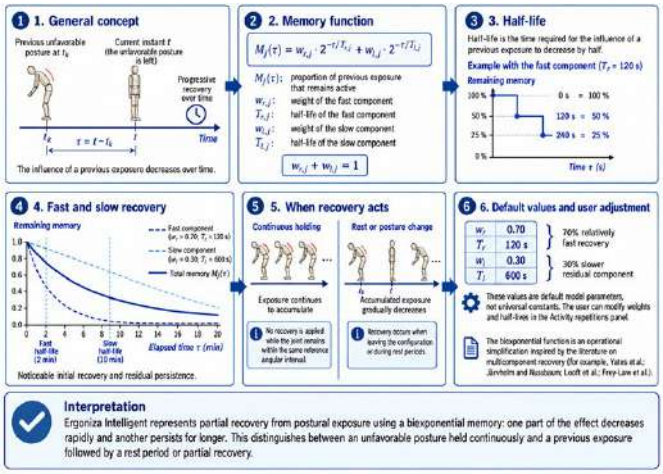
The first term on the right side of Equation 46 represents relatively rapid initial recovery. The second represents slower recovery associated with the persistence of a residual fraction of the prior exposure. The weights of the two components satisfy:

$$w_r + w_l = 1$$

Equation 47

This structure approximates multicomponent behavior without introducing a complex physiological model (Figure 56).

The selection of a memory function with more than one component is supported by the literature on recovery after fatigue. Yates et al.[62] observed that recovery of dynamic muscular endurance did not follow a single phase: it progressed rapidly at first, reached approximately 50% after 2 minutes and 15 seconds, was slightly below 90% after 20 minutes, and was best fitted by a three-component exponential curve. Other studies of localized fatigue models, such as Iridiastadi and Nussbaum[63], also note that recovery is one of the most difficult aspects to represent in intermittent tasks and that its behavior depends on loading conditions.



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Figure 56: Temporal memory and partial recovery of postural discomfort.

In the context of the 3CC-r model, Looft et al.[64] demonstrate the importance of distinguishing recovery during rest intervals. Frey Law et al. [65] show that the relationships among maximum holding time, pain, and perceived exertion are not equivalent in static and dynamic tasks. Taken together, these findings indicate that recovery should not be represented by a single universal behavior.

The biexponential function used by Ergoniza Intelligent is adopted as an operational simplification: it preserves the concept of rapid initial recovery and residual persistence, but avoids presenting the postural-discomfort calculation as an exact physiological model of muscle fatigue.

[62] Yates et al., "Recovery of Dynamic Muscular Endurance."
[63] Iridiastadi and Nussbaum, "Muscle Fatigue and Endurance during Repetitive Intermittent Static Efforts: Development of Prediction Models."
[64] Looft, Herkert, and Frey-Law, "Modification of a Three-Compartment Muscle Fatigue Model to Predict Peak Torque Decline during Intermittent Tasks."
[65] Frey Law et al., "Relationships between Maximum Holding Time and Ratings of Pain and Exertion Differ for Static and Dynamic Tasks."

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The following values are used in the initial Ergoniza Intelligent configuration:

$$w_r = 0.70, T_r = 120\ s$$
$$w_l = 0.30, T_l = 600\ s$$

With these values, 70% of the prior exposure is associated with a relatively fast recovery component with a half-life of 2 minutes (after 120 seconds, it retains 50% of its initial effect; after another 120 seconds, it retains 25%; and so on). The remaining 30% is associated with a slower component with a half-life of 10 minutes. A short break therefore partially reduces cumulative exposure but does not eliminate it completely, while a longer break substantially reduces the influence of prior exposures, although residual memory may remain if the postures are repeated.

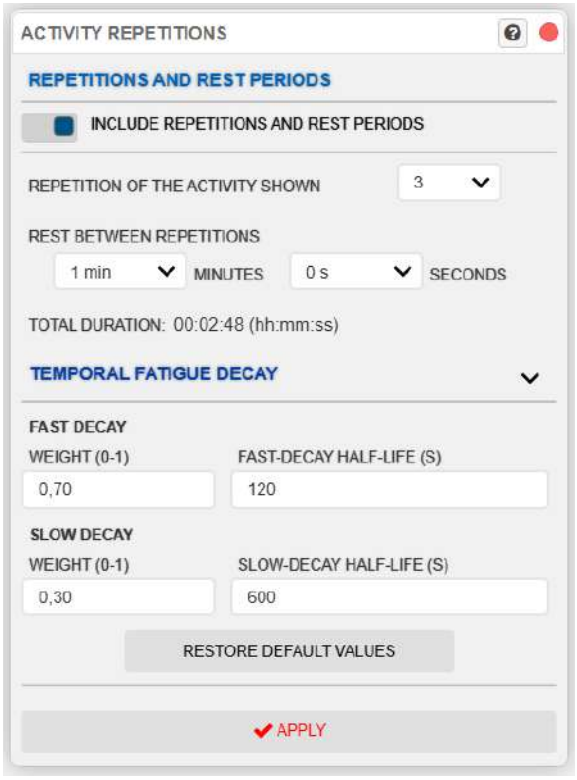


Figure 57: Temporal fatigue decay panel.

Recovery does not apply while a joint remains continuously within the same reference angular interval. In that case, exposure time continues to accumulate. Recovery acts when the joint leaves that configuration or during rest periods between activity repetitions.

The system thus distinguishes between an unfavorable posture sustained continuously and a prior exposure followed by a period of partial recovery. These default values must be interpreted as model parameters, not as universal constants. They were selected to represent appreciable recovery during the first few minutes, but without immediately eliminating the effect of repeated postural exposures. Users can modify them in the **Activity Repetitions** settings (Figure 57) by adjusting both the relative weights of the two components and their half-lives. This dialog can be opened with the **Timing** button on the left sidebar (in Multiple Postures mode) or with the **clock icon** at the bottom of the **Postural Risk** panel shown in Figure 59**.** The **Temporal Fatigue Decay** section is collapsed by default when the Activity Repetitions dialog opens. Click it to expand it.

11.2.4.7 Activity repetitions and rest periods between repetitions

The analyzed posture sequence does not necessarily represent the entire workday or work period; it may instead represent a characteristic cycle that repeats several times. Ergoniza Intelligent can analyze the development of postural discomfort over repeated cycles of a posture sequence (Figure 58).

In this case, the user-defined posture sequence is interpreted as a baseline activity cycle. In the **Repetitions and Rest Periods** section of the **Activity Repetitions** dialog (Figure 57), the user can specify which repetition to represent and how much rest occurs between one repetition and the next.

The system can therefore assess, for example, the postural discomfort corresponding to the first repetition of a task, an intermediate repetition, or a later repetition after prior exposure has accumulated.

If the duration of the baseline cycle is T_c, the rest period between cycles is T_d, and n repetitions are analyzed, the total duration represented by the activity is:

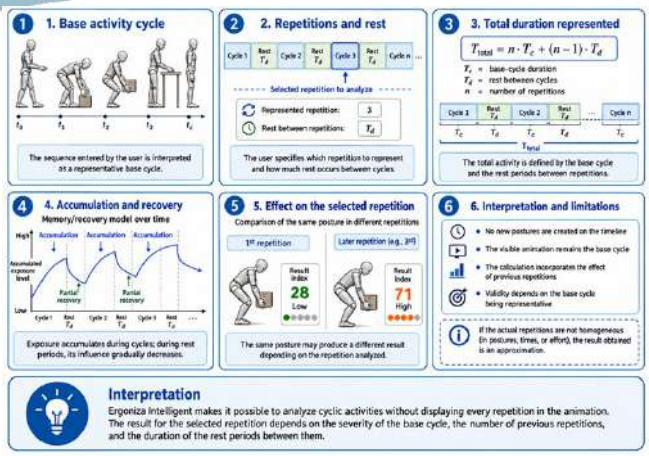
$$T_{total} = n \cdot T_c + (n-1) \cdot T_d$$

Equation 48

The selected repetition is assessed while accounting for exposure accumulated during preceding repetitions. During activity cycles, exposure accumulates in the corresponding reference angular intervals; during rest periods, cumulative exposure is attenuated according to the temporal-memory function described above.

This procedure does not modify the visible animation or create new postures on the timeline. The displayed sequence remains the baseline cycle, but the calculation incorporates the effect of prior repetitions. The same posture within the cycle can therefore produce a different result when analyzed as part of the first repetition or as part of a later repetition.

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Figure 58: Activity repetitions and rest periods between repetitions.

If the first repetition is selected, there is no cumulative prior exposure from earlier cycles. If a later repetition is selected, the result depends on three factors: the postural severity of the baseline cycle, the number of prior repetitions, and the duration of the rest periods. Short rest periods retain more memory of prior exposures, while long rest periods reduce their influence to a greater extent.

The simulation of repetitions must be interpreted as an operational representation of cyclic activities. Its validity depends on the baseline cycle being sufficiently representative of the actual activity. If actual repetitions are not homogeneous, or if there are important variations in pace, pauses, loads, or postures, the result must be interpreted as an approximation of the represented pattern rather than an exact reconstruction of the entire activity.

11.2.4.8 Final calculation of time-relative discomfort

After calculating momentary discomfort $D_{puntual,j}(t)$, absolute discomfort $D_{60,j}(t)$ and equivalent exposure time $T_{eq,j}(t)$, Ergoniza Intelligent obtains final time-relative discomfort for each joint and degree of freedom j.

First, a temporal value associated with equivalent time is calculated:

$$D_{temporal,j}(t) = D_{60,j}(t) \cdot \left(\frac{T_{eq,j}(t)}{60}\right)^{\alpha_j}$$

Equation 49

where α_j is the time exponent associated with the joint and degree of freedom being assessed. This exponent adapts the influence of time to the expected behavior of each body region.

However, the final result is not obtained by applying this expression directly in every case. Ergoniza Intelligent maintains the momentary discomfort of the current posture as the minimum reference. An extreme posture therefore retains its relevance even if it is adopted only briefly or there is no cumulative prior exposure.

The calculation applies the following criteria:

$$D_{final,j}(t) = D_{puntual,j}(t) \text{ if } T_{eq,j}(t) = 0$$

Equation 50

When equivalent time is greater than zero but no more than 60 seconds (the temporal basis of absolute discomfort), the result progresses from momentary discomfort toward absolute discomfort:

$$D_{final,j}(t) = D_{puntual,j}(t) + \left[D_{60,j}(t) - D_{puntual,j}(t)\right] \cdot \left(\frac{T_{eq,j}(t)}{60}\right)^{\alpha_j}$$

Equation 51

This expression applies when $D_{60,j}(t) > D_{puntual,j}(t)$. If momentary discomfort is already equal to or greater than absolute discomfort, the system does not force the result to decrease.

When equivalent time exceeds 60 seconds, the full temporal law is allowed to act, but the final value cannot be lower than momentary discomfort:

$$D_{final,j}(t) = \max\left(D_{puntual,j}(t), D_{temporal,j}(t)\right) \quad \text{if } T_{eq,j}(t) > 60$$

Equation 52

This approach produces a coherent transition among three situations: a posture adopted instantaneously, a posture with an equivalent exposure of 60 seconds or less, and a posture with cumulative exposure greater than the baseline reference time.

The final result therefore combines two distinct effects. First, it preserves the discomfort inherent in the current postural configuration. Second, it increases the result when there is a relevant temporal history of sustaining or repeating similar postures. Time-relative discomfort does not replace momentary discomfort; it uses it as the minimum baseline and modifies it according to cumulative postural exposure.

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11.2.4.9 Accounting for orientation relative to gravity

Orientation relative to gravity for the upper limb is considered in posture sequences using the same general criterion described for single-posture analysis in Section 11.2.2.1. Its purpose is to adjust upper-arm and forearm discomfort when the segment's actual spatial orientation makes the posture more favorable than the relative angle between segments alone would indicate.

In sequence analysis, this correction is calculated at each instant assessed. If **Gravity Adjustment** is enabled (Figure 52), Ergoniza Intelligent checks the orientation of the upper arm and forearm relative to the downward vertical and, when appropriate, partially reduces the corresponding discomfort. This reduction affects discomfort in the current posture and is therefore incorporated before the effect of equivalent exposure time is calculated.

The gravity adjustment does not directly modify equivalent time $T_{eq,j}(t)$, because that quantity represents the temporal history of prior postural exposure. Its effect is introduced into the instantaneous discomfort values of the posture being assessed, that is, into the values that subsequently participate in the temporal combination.

11.2.4.10 Interpretation limits

Time-relative postural discomfort in posture sequences must be interpreted as an ergonomic indicator of equivalent postural exposure. Its purpose is to help identify unfavorable joint configurations that are sustained, repeated, or recur with insufficient rest within a temporally represented activity.

This result must not be interpreted as a clinical prediction of injury, a direct measure of muscle fatigue, or a normative criterion for task acceptability.

The calculation is based on a simplified representation of postural exposure. Although it considers the temporal history of the joint, angular similarity among exposures, the postural relevance of each reference angular interval, and partial recovery, it does not reproduce all the physiological mechanisms involved in fatigue, pain, or individual tolerance.

The result also depends on the quality of the sequence analyzed. If the recorded postures do not adequately represent the actual activity, if the temporal sampling frequency is insufficient, or if important variations are not included in the sequence, the result must be interpreted with caution. For repeated activities, the validity of the calculation also depends on the baseline cycle being representative of the actual repetitions.

Finally, the indicator does not replace biomechanical exertion analysis, lumbar compression analysis, stability assessment, or other postural methods such as REBA or LUBA. It must be used as a complementary measure specifically for assessing the temporal effect of sustaining or repeating unfavorable postural configurations.

11.2.5 Limitations and precautions for use and interpretation

Postural discomfort calculated by Ergoniza Intelligent must be interpreted as an ergonomic indicator of the discomfort associated with a postural configuration and, when applicable, with sustaining or repeating that configuration over time. It must not be used as a clinical, forensic, or regulatory criterion or as a direct predictor of injury or musculoskeletal disorders. Its main purpose is to facilitate comparisons among postures, tasks, or design alternatives by providing a continuous indicator of postural exposure.

11.2.5.1 General limitations of the postural-discomfort indicator

The postural-discomfort calculation is based on psychophysical equations obtained from experimental assessments of comfort in static postures. These equations provide a continuous estimate of the effect of angular deviation on perceived comfort, but they do not reproduce every factor that influences the actual experience of discomfort.

The perception of discomfort depends on individual characteristics such as pain tolerance, age, training, physical condition, preexisting medical conditions, motivation, and familiarity with the task. It can also be modified by environmental factors, the ability to make small postural adjustments, the presence of supports that are not modeled, or the actual variability of the activity. The resulting values must therefore be interpreted as ergonomic estimates rather than exact individualized measurements.

Postural discomfort is also not equivalent to biomechanical exertion. A posture can produce high discomfort because it is close to functional limits or outside comfortable zones, even when the calculated joint moments are low. Similarly, a posture with low angular discomfort may involve substantial exertion if external loads or large moment arms are present. The result must therefore be interpreted together with the analysis of joint exertion, stability, lumbar compression, and, when appropriate, postural methods such as REBA or LUBA.

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11.2.5.2 Limitations of the time-relative calculation for a single posture

When a single posture is analyzed as sustained (Single Posture mode), Ergoniza Intelligent adjusts absolute discomfort to the posture-holding time specified by the user. This adjustment is based on power-law relationships derived from Maximum Endurance Time models. These models originate from studies of static muscle contractions and describe the relationship between relative intensity and holding time to fatigue.

Applying them to postural discomfort requires cautious interpretation. Endurance-time models estimate conditions of muscular tolerance or exhaustion, whereas postural discomfort is a broader perceptual response that is also related to joint position, passive tension, stiffness, pressure, or localized discomfort. The temporal increase in discomfort calculated by Ergoniza Intelligent must therefore be understood as an ergonomic adaptation rather than an exact physiological prediction.

For repetitive postures analyzed in Single Posture mode, the program summarizes repetitiveness using the percentage of the cycle occupied by the exertion and applies an adjustment based on the ACGIH TLV curve. This procedure is an operational approximation and does not explicitly represent the different postures composing the activity. When the task consists of a sequence of distinct postures, analysis in Multiple Postures mode provides a more detailed representation of temporal exposure.

For momentary or unsustained postures, the system does not estimate fatigue or temporal accumulation. The indicator is based on instantaneous proximity to the effective range of motion. It must therefore be interpreted as a preventive indicator of extreme postural deviation, not as an assessment of how long the posture could be sustained.

11.2.5.3 Limitations of the calculation for posture sequences

When a posture sequence is analyzed, Ergoniza Intelligent calculates the effect of prior exposure to similar postural configurations using an equivalent time. This procedure represents the sustaining or repetition of unfavorable postures and partial recovery between exposures. Nevertheless, the model remains a simplification of the actual activity.

The calculation is not a physical dynamic analysis. It does not incorporate inertia, acceleration, or forces resulting from movement. The sequence is interpreted as a temporal succession of postures, and the result expresses equivalent postural exposure rather than dynamic mechanical load.

The result depends on the quality and representativeness of the sequence analyzed. If the recorded postures do not adequately represent the actual activity, the temporal sampling frequency is insufficient, rapid movements are not captured, or the actual task has important variations that are not included in the sequence, the result must be interpreted with caution.

For repeated activities, the repetition simulation assumes that the baseline cycle entered by the user is representative of the actual cycles. If the actual repetitions are not homogeneous, or if there are important variations in pace, pauses, loads, postures, or movement strategy, the result represents an approximation of the described pattern rather than an exact reconstruction of the entire activity.

The temporal-memory and partial-recovery parameters used in the calculation are operational model parameters. Although they are informed by the literature on multicomponent recovery and intermittent tasks, they must not be interpreted as universal physiological constants. Their purpose is to represent operationally that a recent exposure has a greater influence than an older one and that rest periods progressively reduce cumulative exposure without necessarily eliminating it immediately.

Time-relative postural discomfort in posture sequences must therefore be used as a comparative analysis tool. It is particularly useful for locating moments of high postural exposure within an activity, comparing alternative designs, or assessing the effect of introducing breaks, modifying the movement sequence, or reducing the time spent in unfavorable joint configurations.

11.2.6 Visualizing discomfort on the model

Ergoniza Intelligent displays postural discomfort in the Postural Discomfort section of the Postural Load panel (Figure 59).

When discomfort reaches or exceeds 100% (or the threshold configured by the user for the Risk level), the chart is shown in red. When the percentage exceeds 85% (or the threshold configured by the user for the Warning level), the corresponding chart is shown in orange.

The information can also be displayed directly on the model using the Visualization on the Model button, identified by an eye icon, in the Postural Discomfort section. When enabled, the model's body segments are colored using the same color code.

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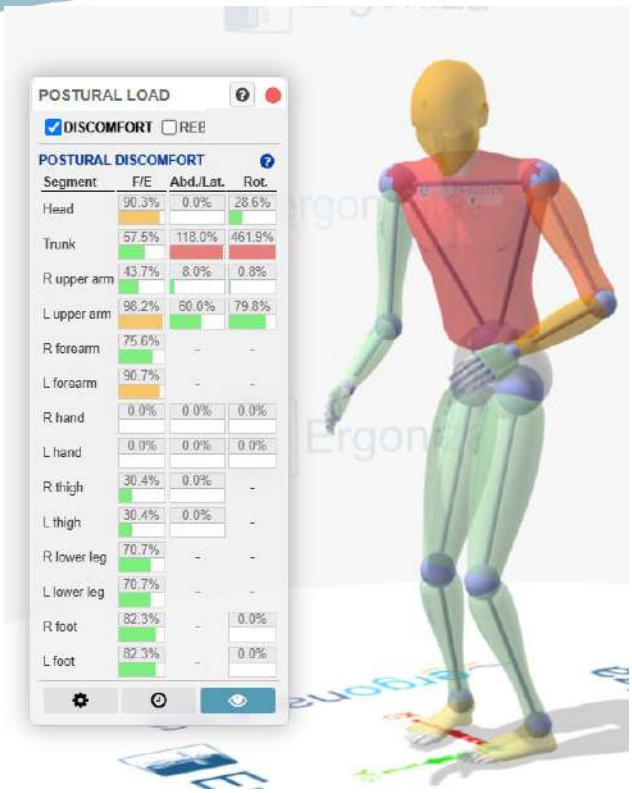


Figure 59: Postural-discomfort visualization on the model.

11.3 Assessing posture-related risk using the REBA and LUBA methods

The REBA and LUBA methods use conceptually different approaches to assess the risk associated with adopted postures. REBA (Rapid Entire Body Assessment) is based on the level of exertion required to maintain a posture and considers factors such as handled loads, number of repetitions, applied forces, and postures adopted throughout the body. LUBA (Loading on the Upper Body Assessment), in contrast, focuses on the subjective assessment of perceived discomfort in different torso and upper-extremity segments.

This difference means that REBA provides a more direct estimate of objective physical risk, whereas LUBA more closely reflects the discomfort reported by people performing the activity. The latter can be particularly useful where perceived comfort affects task sustainability or the progressive development of musculoskeletal disorders.

LUBA has several limitations. First, it does not directly consider the handled load or the duration of a sustained posture. It also excludes the lower extremities from its analysis, making it unsuitable for assessing tasks in which the legs or feet are substantially involved, such as squatting, kneeling, or prolonged standing. Nevertheless, the LUBA approach can identify postures that may be uncomfortable or unsustainable for workers over the long term even when other methods classify them as biomechanically acceptable.

In general, LUBA should be used to assess the comfort of static postures that must be sustained over time rather than dynamic postures that change continuously.

11.3.1 Assessing postural risk using the REBA method

Ergoniza Intelligent assesses postural load using the REBA method[66]. This whole-body observational method analyzes the positions of the trunk, neck, legs, upper arms, forearms, and wrists and combines them with factors related to the applied load or force, coupling, and type of activity. The result is a score associated with a risk and action level that indicates how urgently corrective measures should be considered.

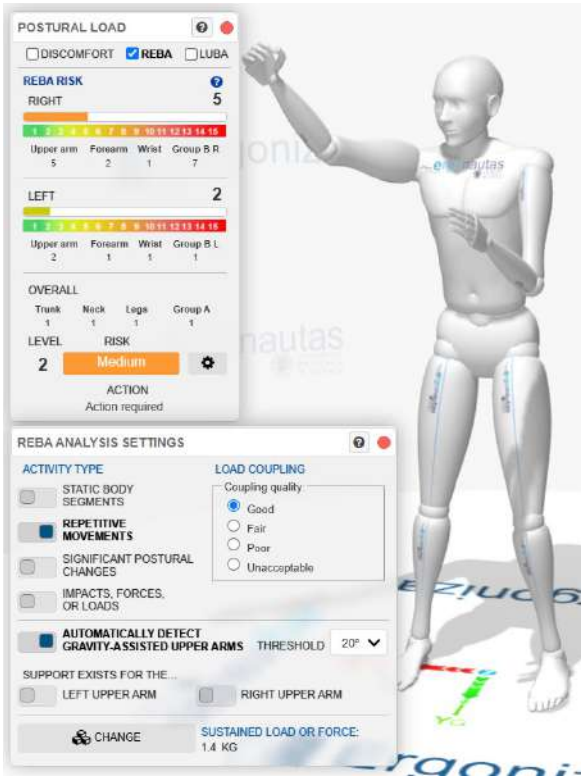


Figure 60: Postural-load assessment using the REBA method in Ergoniza Intelligent.

Ergoniza Intelligent automatically calculates the score for each body segment from the model posture, as well as the partial scores for Groups A and B defined by the method. It then

[66] Hignett, S. & McAtamney, L., (2000). REBA: Rapid Entire Body Assessment. Applied Ergonomics, 31, pp.201-205.

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obtains a final score for each side of the body and displays the risk and action level for the less favorable side as the overall result in the REBA section of the Postural Load panel (Figure 60).

The result obtained through REBA analysis depends on parameters such as body segments that remain static, repetitive movements, and coupling quality. These parameters can be defined in the *REBA Analysis Settings* dialog. To open it, use the gear button in the **REBA** section.

When assessing the upper arm, REBA reduces the score if the arm is supported or if its posture is assisted by gravity. This condition allows different scores to be assigned to situations in which the upper arm adopts the same angular position but is either supported or assisted by gravity.

Ergoniza Intelligent can automatically detect whether an upper arm is favorably oriented relative to gravity. This avoids requiring the user to mark the condition manually when the arm's orientation allows it to be inferred geometrically. The system analyzes the spatial direction of the upper arm, defined by the segment between the shoulder and elbow, and compares it with the downward vertical. If the upper arm is sufficiently aligned with the direction of gravity, the system interprets the posture as gravity-assisted and automatically applies the REBA score reduction for the upper arm. To use this automatic detection, enable Automatically Detect Gravity-Assisted Upper Arms in the **REBA Analysis Settings** dialog (Figure 60).

When automatic detection is enabled, the manual upper-arm checkboxes indicate only that support exists. Whether the posture is gravity-assisted is determined automatically from the upper arm's orientation. When automatic detection is disabled, the manual checkboxes regain their general REBA meaning and can indicate either that support exists or that the upper-arm posture is assisted by gravity.

The user can also define the angular threshold used for automatic detection. This threshold specifies how close the upper arm must be to the downward vertical for Ergoniza Intelligent to consider the posture gravity-assisted. A low value makes detection stricter and applies it only when the upper arm hangs almost vertically. A higher value makes detection more permissive and allows it to apply to positions somewhat farther from vertical.

This automation does not modify the REBA method or its general scoring rules. Its purpose is to assist the assessor in identifying one of the conditions considered by the method. As with any ergonomic assessment, the result must be reviewed in light of the actual task, the presence of supports, the handled loads, and the analyst's professional judgment.

11.3.2 Assessing risk from unsuitable postures using the LUBA method

The LUBA method [67] (*Loading on the Upper Body Assessment*) is a technique for assessing static postural load on the upper body. It assigns relative discomfort scores to movements of the hand and wrist, elbow, shoulder, neck, and torso joints according to their angular deviation from the neutral position. Each movement is classified into angular intervals or postural classes associated with an experimentally obtained discomfort score, as shown in the Joint-Motion Discomfort section of Figure 61. The sum of the scores for a posture constitutes **the Postural Load Index**), which is inversely related to Maximum Holding Time (MHT): the higher the index, the shorter the time for which the posture can be expected to be sustained without a substantial increase in discomfort. Ergoniza Intelligent performs this calculation independently for the right and left sides of the body.

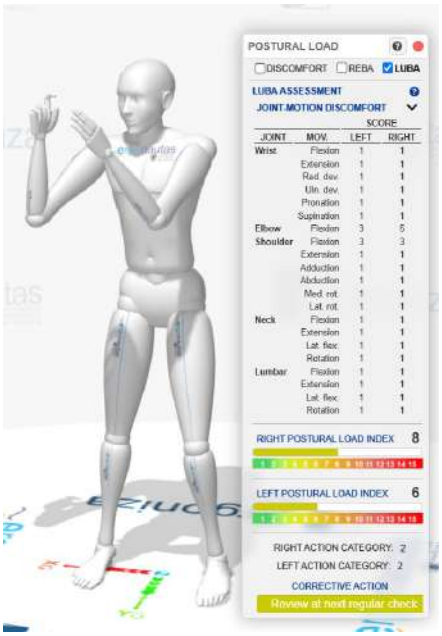


Figure 61: Postural-load assessment using the LUBA method in Ergoniza Intelligent

Based on the relationship between the Postural Load Index and Maximum Holding Time, LUBA defines four action categories (I, II, III, and IV) indicating the need and urgency for corrective measures. Ergoniza Intelligent calculates one category for the

[67] Kee, D. & Karwowski, W., (2001). LUBA: an assessment technique for postural loading on the upper body based on joint motion discomfort and maximum holding time. Applied Ergonomics, 32, pp. 357-366 <https://doi.org/10.1016/S0003-6870(01)00006-0>

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right side and another for the left and presents the action corresponding to the less favorable result as the overall action.

12 LUMBAR COMPRESSION AND SHEAR RISK ANALYSIS

External loads and the weight of body segments above the lumbar region generate moments that must be balanced by muscle forces and joint reactions. Consequently, the intervertebral discs withstand compression and shear forces that can increase during load handling, especially when the load is heavy, held away from the body, or the trunk adopts unfavorable postures. Estimating these forces is therefore used as a biomechanical indicator of lumbar demand. Ergoniza Intelligent performs this calculation at the L4/L5 intervertebral level.

In its 1981 manual-handling guide, the U.S. National Institute for Occupational Safety and Health (NIOSH)[68] established two reference values for estimated compression force at L5/S1: an Action Limit (*Action Limit*, AL) of 3400 N and a Maximum Permissible Limit (*Maximum Permissible Limit*, MPL) of 6400 N. The revised NIOSH equation subsequently retained 3400 N as a biomechanical criterion associated with increased risk of low-back injury.

These values were established for L5/S1 and for the ergonomic assessment of tasks. They must therefore not be interpreted as individual injury limits or as a precise boundary between a safe and an injurious situation. Ergoniza Intelligent uses them as operational references to classify the compression force estimated by its model at L4/L5. Applying them to a different, albeit adjacent, lumbar level is an approximation that must be considered when interpreting the result.

Subsequent research examined how age and sex influence the load-bearing capacity of the spine, particularly at L5/S1[69]. Aging causes progressive degeneration of the intervertebral discs and loss of mechanical strength in the vertebrae and supporting tissues. Biomechanical studies show that the compressive load capacity of the discs decreases substantially after age 40 or 50. Some authors therefore propose more conservative compression limits for older populations. In general, women also have less bone and muscle mass than men, which may reduce their capacity to withstand high compressive loads. Studies suggest that women's intervertebral discs have smaller diameters and lower absolute resistance to compressive loads.

Although the NIOSH limit of 3400 N remains a general reference for mixed populations, more advanced ergonomic applications such as Ergoniza Intelligent can adjust safety criteria for age and sex. Table 6 shows the recommended L5/S1 intervertebral compression-force limits used by Ergoniza Intelligent according to age and sex. These values are based on the studies cited above and are calculated automatically when the corresponding model data are entered.

Group	Recommended limit	Notes	Ergoniza Intelligent limit
Young adults (20-40 years)	3400 N	Original general limit.	3400 N
Middle-aged adults (40-50 years)	3000-3200 N	Moderate intervertebral-disc degeneration.	3100 N
Older adults (>50 years)	2500-2800 N	Greater disc degeneration and injury risk.	2650 N
Young women (20-40 years)	3000-3200 N	Lower average structural strength is assumed.	3100 N
Older women (>50 years)	2500 N	A conservative approach is strongly recommended.	2500 N

Table 6: Operational compression thresholds used by Ergoniza Intelligent according to sex and age.

Ergoniza Intelligent also uses a conservative operational threshold of 500 N for **resultant shear** at L4/L5. This resultant combines the anteroposterior and lateral components. The value originates from classic biomechanical studies[70] that examined the resistance of ligamentous and disc structures to

[68] NIOSH National Institute for Occupational Safety and Health, *Work Practices Guide for Manual Lifting*, 1981; T R Waters, V Putz-Anderson, and A Garg, *Applications Manual for the Revised NIOSH Lifting Equation* (Cincinnati, 1994).

[69] William S. Marras, Kermit G. Davis, and Michael Jorgersen, "Spine Loading as a Function of Gender," *Spine* 27, no. 22 (2002): 2514–20, doi:10.1097/00007632-200211150-00017; William S. Marras, Kermit G. Davis, and Michael Jorgensen, "Gender Influences on Spine Loads during Complex Lifting.," *The Spine Journal : Official Journal of the North American Spine Society* 3, no. 2 (2003): 93–99, doi:10.1016/S1529-9430(02)00570-3; Sean Gallagher and William S. Marras, "Tolerance of the Lumbar Spine to Shear: A Review and Recommended Exposure Limits," *Clinical Biomechanics* 27, no. 10 (2012): 973–78, doi:10.1016/j.clinbiomech.2012.08.009; Don B.J. Chaffin, Gunnar B. Andersson, and Bernard J. Martin, *Occupational Biomechanics, 4th Edition.* (Wiley and Sons, 2006); P. Brinckmann, M. Biggemann, and D. Hilweg, "Fatigue Fracture of Human Lumbar Vertebrae," *Clinical Biomechanics*, 1988, doi:10.1016/S0268-0033(88)80001-9.

[70] M. A. Adams and W. C. Hutton, "The Mechanical Function of the Lumbar Apophyseal Joints," *Spine* 8, no. 3 (1983): 327–30, doi:10.1097/00007632-198304000-00017; D. B.J. Chaffin, G. B. J. Andersson, and B.J. Martin, *Occupational Biomechanics*, Fourth (Hoboken, NJ: Jhon Wiley & Sons Inc., 2006).

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shear stress. Later reviews proposed values of 1000 N for occasional exposures and 700 N for repetitive exposures; the 500 N value used by the program must therefore be understood as a conservative screening criterion. Exceeding it indicates that the task should be reviewed, but does not by itself predict injury.

12.1 Calculating compression and shear in the L4/L5 intervertebral disc

The compression force acting on the lumbar intervertebral discs does not depend solely on the weight of an external load. The weight of body segments and sustained loads generates moments in the lumbar region that must be balanced by the trunk muscles. These include the posterior erector spinae and several abdominal-wall muscles. The muscle forces required to maintain equilibrium can add a substantial compression component to the disc, so the calculation must jointly consider external loads, body weight, and estimated muscle activity.

Intra-abdominal pressure[71] also contributes to the mechanical equilibrium of the lumbar region. Coordinated contraction of the abdominal wall, diaphragm, and pelvic floor increases pressure within the abdominal cavity and generates a force whose moment can help balance the flexion moment. In the model, this contribution reduces part of the moment that the trunk muscles must counteract. However, intra-abdominal pressure and muscle activation are related, so its effect on disc compression must not be interpreted as direct unloading independent of the other forces.

Decomposing forces into axial-compression and shear components requires an estimate of the spatial orientation of the L4/L5 disc in the analyzed posture. Ergoniza Intelligent calculates this orientation from the configuration of the trunk and lower extremities and projects the muscle forces, intra-abdominal pressure, and resultant external loads onto the disc's local coordinate system.

The calculation therefore requires the resultant force and moment of the body segments above L4/L5 and the external loads, an estimate of the contribution of intra-abdominal pressure and the trunk muscles, and the local orientation of the disc. These quantities yield the axial-compression force and the anteroposterior and lateral shear components.

Ergoniza Intelligent performs these calculations using a three-dimensional model with ten trunk muscles grouped into five bilateral pairs. The calculation is performed at the L4/L5 intervertebral level, for which anatomical information is available on muscle areas, lines of action, and moment arms used by this type of model.

Figure 62 illustrates the procedure. First, Ergoniza Intelligent estimates the orientation of the transverse plane of the L4/L5 intervertebral disc from the posture of the trunk and lower extremities[72]. This plane and its local reference system are shown in blue in the figure.

The objective is to calculate the disc compression force (**C** in Figure 62) and the anteroposterior (**Cor y**) and lateral (**Cor x**) shear forces. These three components form the reaction required to balance the resultant of the other forces in the disc's local coordinate system. The vector sum of moments about L4/L5 must also equal zero.

The forces considered are the force associated with intra-abdominal pressure (**Fa**), the weight of the segments above L4/L5 and sustained loads (**P**), the contraction forces of the ten trunk muscles (**Xd, Xi, Id, Ii, Ed, Ei, Ld, Li, Rd**, and **Ri**), and the three components of the disc reaction.

The resultant **P** and the moment it generates about L4/L5 are obtained through the joint-load analysis described in Section 8. Force **Fa** and its moment are calculated using the procedure described in Section 12.1.1.

The model includes five pairs of trunk muscles: *Latissimus dorsi, Erector spinae, Rectus abdominis, Obliquus externus*, and *Obliquus internus*. A cross-sectional area, line of action, and moment arm about L4/L5 are defined for each muscle. Ergoniza Intelligent uses separate parameter sets for men and women derived from the wide range of studies addressing abdominal-muscle geometry[73].

[71] Jacek Cholewicki, Krishna Juluru, and Stuart M. McGill, "Intra-Abdominal Pressure Mechanism for Stabilizing the Lumbar Spine," *Journal of Biomechanics* 32, no. 1 (1999): 13–17, doi:10.1016/S0021-9290(98)00129-8.

[72] Anderson, Chaffin, and Herrin, "A Study of Lumbosacral Orientation under Varied Static Loads."

[73] Michael E. Gerling and Stephen H.M. Brown, "Architectural Analysis and Predicted Functional Capability of the Human Latissimus Dorsi Muscle," *Journal of Anatomy* 223, no. 2 (2013): 112–22, doi:10.1111/joa.12074; J. G. Reid and P. A. Costigan, "Geometry of Adult Rectus Abdominis and Erector Spinae Muscles," *Journal of Orthopaedic and Sports Physical Therapy*, 1985, doi:10.2519/jospt.1985.6.5.278; Hyung Woo Koh, Sung Hyoun Cho, and Cheol Yong Kim, "Comparison of the Effects of Hollowing and Bracing Exercises on Cross-Sectional Areas of Abdominal Muscles in Middle-Aged Women," *Journal of Physical Therapy Science* 26, no. 2 (2014): 295–99, doi:10.1589/jpts.26.295; S. M. McGill, N. Patt, and R. W. Norman, "Measurement of the Trunk Musculature of Active Males Using CT Scan Radiography: Implications for Force and Moment Generating Capacity about the L4 L5 Joint," *Journal of Biomechanics* 21, no. 4 (1988): 329–41, doi:10.1016/0021-9290(88)90262-X; J. Gavin Reid, P. A. Costigan, and W. Comrie,

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"Prediction of Trunk Muscle Areas and Moment Arms by Use of Anthropometric Measures," *Spine* 12, no. 3 (1987): 273–75, doi:10.1097/00007632-198704000-00015; Gunnar Németh and Hans Ohlsén, "Moment Arm Lengths of Trunk Muscles to the Lumbosacral Joint Obtained in Vivo with Computed Tomography," *Spine* 11, no. 2 (1986): 158–60, doi:10.1097/00007632-198603000-00011; Shrawan Kumar, "Moment Arms of Spinal Musculature Determined from CT Scans," *Clinical Biomechanics* 3, no. 3 (1988): 137–44, doi:10.1016/0268-0033(88)90059-9; S. Kumar, Don B. Chaffin, and Mark Redfern, "Isometric and Isokinetic Back and Arm Lifting Strengths: Device and Measurement," *Journal of Biomechanics* 21, no. 1 (1988): 35–44, doi:10.1016/0021-9290(88)90189-3; Chaffin et al., "Lumbar Muscle Size and Locations from CT Scans of 96 Women of Age 40 to 63 Years."

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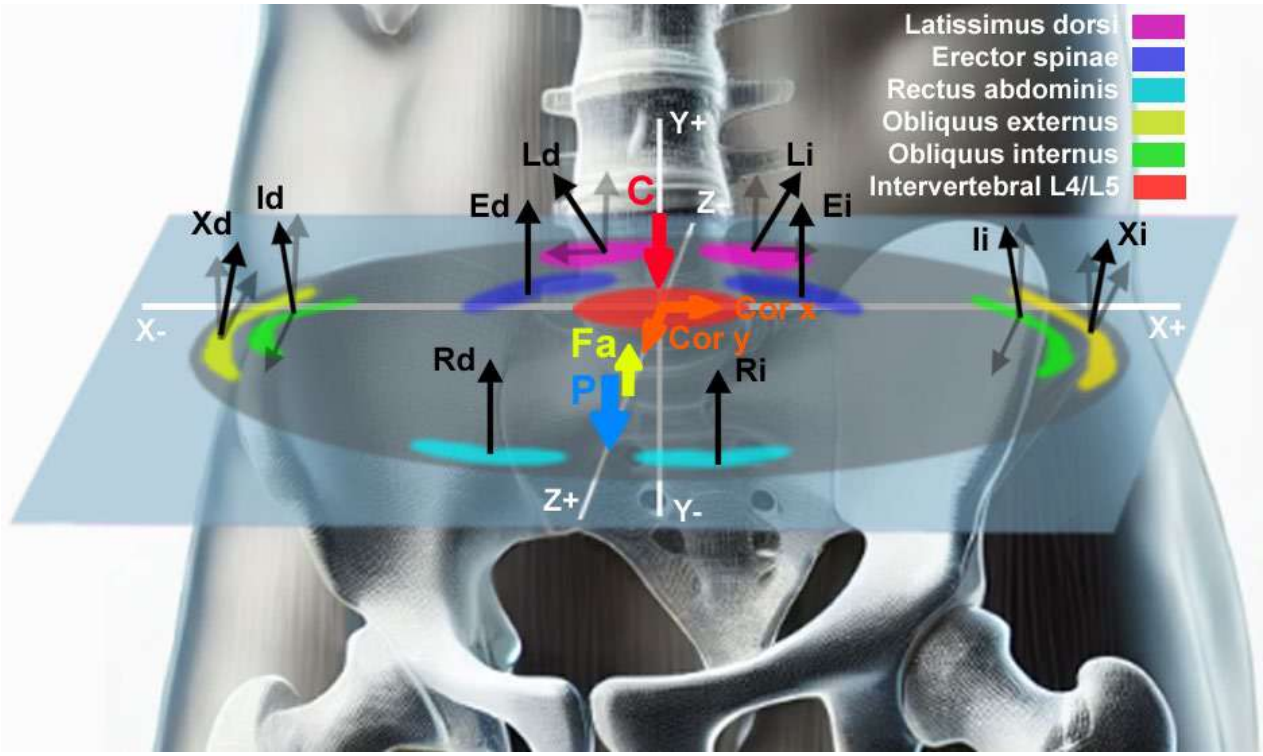


Figure 62: Three-dimensional model for calculating L4/L5 intervertebral-disc compression.

When the model is seated on a chair and **Backrest** is enabled, Ergoniza Intelligent applies the simplification described in Section 7.7.6. If the local flexion-extension component of the moment at L4/L5 corresponds to flexion, that component is set to zero before muscle activity is calculated. The lateral-flexion and rotation components are retained and the external reaction forces are not modified because neither the geometry nor the application point of the backrest reaction is modeled. The program recalculates muscle activity, compression, and shear forces from the corrected moment. Compression therefore does not necessarily become zero or decrease in every case.

Determining muscle forces is a statically indeterminate problem because there are more unknown forces than independent equilibrium equations. Ergoniza Intelligent solves it using two successive linear-programming problems, described in Section 12.1.2. The force associated with intra-abdominal pressure and its moment arm are described first.

12.1.1 Force associated with intra-abdominal pressure

In the model, intra-abdominal pressure is represented by an equivalent force $\mathbf{F_a}$ whose line of action lies anterior to the lumbar spine. The moment produced by this force helps balance the flexion-extension component of the moment at L4/L5 and therefore modifies the moment that the trunk muscles must counteract.

The force generated by intra-abdominal pressure can be estimated [74] using Equation 53, where F_A is the force generated by intra-abdominal pressure in newtons, θ_t is the angle in degrees between the trunk axis and the extension of the thigh axis (as shown in Figure 63), and $M_{L5/S1}$ is the component perpendicular to the sagittal plane of the moment caused by handled loads and the weight of body segments above the joint (in N·m). To determine F_A, the model uses an average diaphragm area [75] of 465 cm^2.

[74] Chaffin, Andersson, and Martin, *Occupational Biomechanics, 4th Edition.*; J. M. Morris, D. B. Lucas, and B. Bresler, "Role of the Trunk in Stability of the Spine," *The Journal of Bone & Joint Surgery* 43A (1961): 327–51, doi:10.2106/00004623-196143030-00001.

[75] Morris, Lucas, and Bresler, "Role of the Trunk in Stability of the Spine."

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$$F_A = 6.199473 \cdot 10^{-4} \ (43 \ - \ 0.36 \ \theta_t \ )(M_{L5/S1})^{1.8}$$

Equation 53

Determining the moment generated by this force about the lumbar joint requires its moment arm (D_{FA} in Figure 63). The moment arm ranges from approximately 7 to 15 cm[76] depending on the sine of angle θ_t; its value in meters can be estimated using Equation 54.

$$D_{FA} = 0.066 + 0.082 \ ABS(SEN(\theta_t \cdot \ \pi \ / \ 180))$$

Equation 54

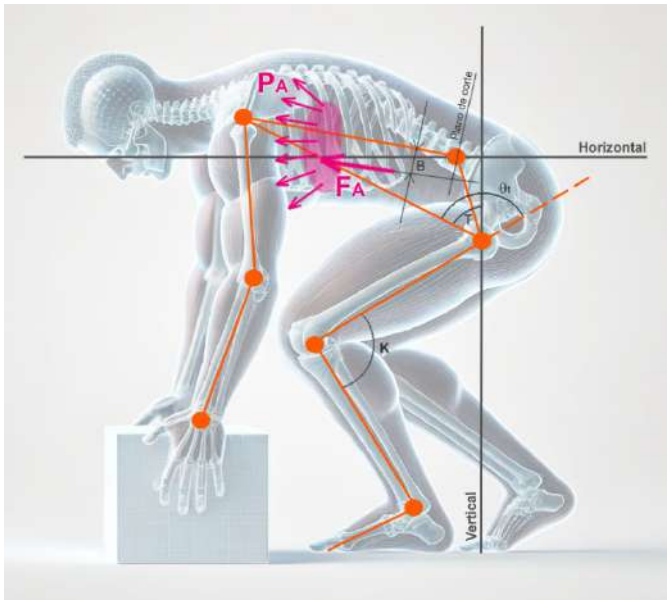


Figure 63: Effect of intra-abdominal pressure.

Once the force generated by intra-abdominal pressure F_A and its moment arm D_{FA} are known, their product gives the moment associated with intra-abdominal pressure, which is incorporated into the flexion-extension component of equilibrium at L4/L5.

12.1.2 Iterative model for calculating trunk-muscle forces

During load lifting, especially asymmetric lifting, trunk muscles are activated to help sustain the posture. This muscular redundancy results in more unknown forces than force- and moment-equilibrium equations, making the problem statically indeterminate. The three-dimensional model used by Ergoniza Intelligent to calculate compression forces in the L4/L5 intervertebral disc (Section 12.1) contains 13 unknown forces—those corresponding to muscle contractions and intervertebral-disc reactions—but only 6 equations.

To solve this system, constraints are introduced regarding how the muscles act[77]. These constraints establish that muscle-contraction intensities—force per unit cross-sectional area—and intervertebral-disc compression force are minimized during exertion. This is equivalent to assuming that, during load lifting, the central nervous system attempts to protect the muscles and intervertebral discs by minimizing the stresses imposed on them[78].

Because both compressive forces and muscle intensities must be minimized, the problem can be solved using a two-stage linear-programming optimization procedure[79]. First, maximum muscle intensity is minimized while satisfying moment-equilibrium conditions. The muscle forces that minimize the muscular contribution to vertebral compression are then determined while respecting the intensity limits.

The **first stage** minimizes **the maximum contraction intensity** among all muscles. For each muscle *i*, $\mathbf{F_i}$ represents the magnitude of its contraction force. From the position and line of action of the muscle, the model determines the moment generated about L4/L5 by a unit force from that muscle. This vector coefficient is represented by $\mathbf{c_i}$ and depends on the geometry of the analyzed posture. The moment generated by the muscle is therefore $\mathbf{c_iF_i}$, expressed in N·m, and varies linearly with its contraction force. At equilibrium:

$$\sum_{i=1}^{10} c_i F_i = -M_{ext,corr}$$

where $\mathbf{M_{ext,corr}}$ is the external moment that the muscles must balance, expressed in the L4/L5 local coordinate system after incorporating the effect of intra-abdominal pressure and, when applicable, the backrest simplification. The preceding equation actually represents three independent equations, one for each axis of the L4/L5 local coordinate system.

[76] D.B. Chaffin, "On the Validity of Biomechanical Models of the Low-Back for Weight Lifting Anlaysis," *Amer. Soc. Mech. Eng Proc.*, nos. 75-WA-B10-1 (1975).
[77] A. Schultz, G. Andersson, and R. Ortengren, "Loads on the Lumbar Spine. Validation of a Biomechanical Analysis by Measurements of Intradiscal Pressures and Myoelectric Signals," *Journal of Bone and Joint Surgery - Series A* 64, no. 5 (1982): 713–20, doi:10.2106/00004623-198264050-00008.
[78] Chaffin, Andersson, and Martin, *Occupational Biomechanics*.
[79] James C. Bean, Don B. Chaffin, and Albert B. Schultz, "Biomechanical Model Calculation of Muscle Contraction Forces: A Double Linear Programming Method," *Journal of Biomechanics* 21, no. 1 (1988): 59–66, doi:10.1016/0021-9290(88)90192-3.

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The linear-programming problem is formulated as:

$$\min \lambda$$

subject to:

$$\sum_{i=1}^{10} c_i F_i = -M_{ext,corr}$$

$$F_i \geq 0 \quad i = 1, \ldots, 10$$

$$\frac{F_i}{A_i} \leq \lambda \quad i = 1, \ldots, 10$$

where **λ** is the greatest muscle intensity, expressed in N/cm^2, and $\mathbf{A_i}$ is the physiological cross-sectional area of muscle **i**, in cm^2.

In the **second stage**, once **λ** has been determined, a second linear-programming problem is solved to **minimize the total compressive force on the intervertebral disc**. The second problem is formulated as:

$$\min \sum_i F_{i,}$$

subject to the same equilibrium equations, $F_i \geq 0$, and:

$$\frac{F_i}{A_i} \leq \lambda$$

Once the muscle forces have been determined, they are projected onto the L4/L5 local coordinate system. Their components perpendicular to the disc plane contribute to axial compression, while the components within the plane contribute to shear.

After obtaining the muscle forces, the disc reaction is calculated by imposing force equilibrium. Projecting this reaction onto the axis normal to the L4/L5 plane gives axial compression, while its projections onto the anteroposterior and lateral axes give the two shear components.

Ergoniza Intelligent displays the calculation results in a window like the one shown in Figure 64. In this example, estimated axial compression is **2609.5 N**, below the operational threshold of **3400 N** applicable to the selected model. The shear components are **289.4 N** in the anteroposterior direction and **0.0 N** in the lateral direction; their resultant is also below the **500 N** threshold. The displayed diagnosis is therefore **Acceptable**.

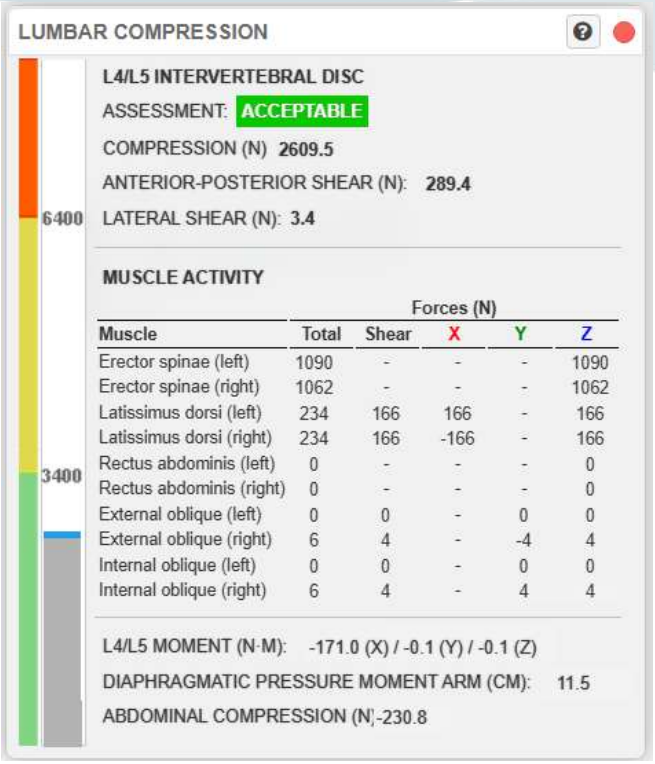


Figure 64: Results of the lumbar-compression calculation for the L4/L5 intervertebral disc in Ergoniza Intelligent

The **Muscle Activity** section presents, for each of the ten muscles considered, the calculated total force and its components in the L4/L5 local coordinate system: **X**, lateral; **Y**, anteroposterior; and **Z**, axial or normal to the disc plane. The window also shows the three components of the moment at L4/L5, the moment arm of intra-abdominal pressure, and its associated equivalent force.

Ergoniza Intelligent verifies that both optimization problems are feasible, that muscle forces are nonnegative, and that moment-equilibrium residuals remain within the established tolerance. If no usable solution is obtained, the window displays **Unable to Calculate**.

13 LOAD LIFTING ANALYSIS

Ergoniza Intelligent includes a specific mode for analyzing tasks in which a worker manually raises or lowers a load between an origin and a destination. The analysis considers lifting geometry and exposure conditions through an implementation of the Revised NIOSH Lifting Equation.

The **Load Lifting mode** can be activated with the corresponding button at the top center of the interface (Figure 1). Unlike single- or multiple-posture analysis, the lifting module does not use an isolated body posture or a generic temporal sequence as its main unit of analysis. Instead, it analyzes a **lifting activity** that may comprise several tasks.

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The assessment has two distinct dimensions. A **lifting task** represents specific handling conditions: weight, frequency, coupling, handling method, and origin and destination geometry. A **lifting period** represents the interval during which the task profile is performed before recovery.

An assessment can contain one or more tasks and, independently, one or more periods. All periods currently use a shared profile: the same tasks are performed at the same frequencies in every period. Periods are not additional NIOSH tasks; they describe when that profile is performed.

Temporal organization can be defined as one lifting period, a regular pattern of periods, or a custom sequence. The entered durations and recovery times determine the evaluation blocks and the duration category applicable to the frequency multiplier.

In the Load Lifting mode interface, users can define general analysis parameters, create one or more lifting tasks, adjust the specific conditions of each task, and view the resulting overall and partial results, as shown in Figure 65. Origin and destination results are displayed separately because conditions may differ at the two points of the movement.

In a single-task analysis, each block is assessed using the task's Lifting Index. In a multitask analysis, each block is assessed using the Composite Lifting Index for the task set. All blocks are calculated and compared. The block producing the least favorable index is identified as the governing block and establishes the representative overall assessment result. Results for the other blocks are retained for review but are neither added nor averaged.



Figure 65: Ergoniza Intelligent Load Lifting mode.

13.1 Purpose, scope, and basis of the analysis

Load Lifting mode in Ergoniza Intelligent assesses manual-handling tasks in which a person raises or lowers a load between an origin and a destination. It analyzes a task defined by its geometric, temporal, and operating conditions rather than an isolated posture.

The analysis is based on the **Revised NIOSH Lifting Equation**[80]. Its main result is the **Recommended Weight Limit**, or **RWL** (*Recommended Weight Limit*), a reference weight for the conditions studied. It must not be interpreted as an individual limit guaranteeing the absence of injury. The RWL is used to calculate the **Lifting Index**, or **LI**, which expresses the ratio between the actual load weight and the RWL:

$$LI = L / RWL$$

Equation 55

where **L** is the load weight.

NIOSH recommends an LI of 1 or less. A value above 1 indicates that the load exceeds the Recommended Weight Limit for the analyzed conditions and that the task should be

[80] National Institute for Occupational Safety and Health, *Work Practices Guide for Manual Lifting*; NIOSH, "Applications Manual for the Revised NIOSH Lifting Equation" (DHHS (NIOSH) Publication No. 94-110, 2021), doi:10.26616/NIOSHPUB94110revised092021.

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reviewed. The index is a relative estimate of physical demand; it does not express an individual's probability of injury or mean that a task with LI ≤ 1 is risk-free.

The RWL is calculated using Equation 56, which starts from a Load Constant (LC) and applies several multipliers related to lifting characteristics. These factors include the horizontal distance of the load from the body, vertical height, vertical travel distance between origin and destination, lifting asymmetry, handling frequency, task duration, and coupling quality.

$$RWL = LC \cdot HM \cdot VM \cdot DM \cdot AM \cdot FM \cdot CM$$

Equation 56

In this expression, **LC** is the Load Constant; **HM**, the Horizontal Multiplier; **VM**, the Vertical Multiplier; **DM**, the Distance Multiplier for vertical travel; **AM**, the Asymmetry Multiplier; **FM**, the Frequency Multiplier accounting for frequency, duration, and recovery; and **CM**, the Coupling Multiplier. All multipliers have a maximum value of 1 and reduce the RWL as conditions depart from the reference conditions.

Ergoniza Intelligent obtains the geometric parameters and calculates results for both the origin and destination. However, the destination contributes to the task result only when significant control of the load is required at that point. When such control exists, the program compares both endpoints and adopts the less favorable condition. Otherwise, the task result is determined from the origin, although destination data and results are retained to describe and review the movement. Significant control exists when the load must be regrasped, held, stabilized, guided, or positioned precisely at the destination.

The Revised NIOSH Lifting Equation is not a general assessment of pushing, pulling, carrying, seated or kneeling lifting, very rapid movements, unstable loads, or other forms of handling outside its scope. Ergoniza Intelligent also includes extensions that are not part of the conventional formulation, such as anthropometric adjustment, one-handed handling, and a configurable Load Constant. These extensions must be considered when interpreting the results.

13.2 Defining lifting tasks, general parameters, and anthropometric adjustment

In Load Lifting mode, the analysis is organized around one or more **lifting tasks**. An analysis can contain from one to forty tasks. Each task represents an operation performed under a specific set of conditions. If the same activity includes lifts with different weights, frequencies, positions, couplings, or handling methods, they must be entered as separate tasks.

The main panel shows the tasks created and the results for each one (Figure 66). From this panel, users can select the active task, view its results, and access its configuration options.

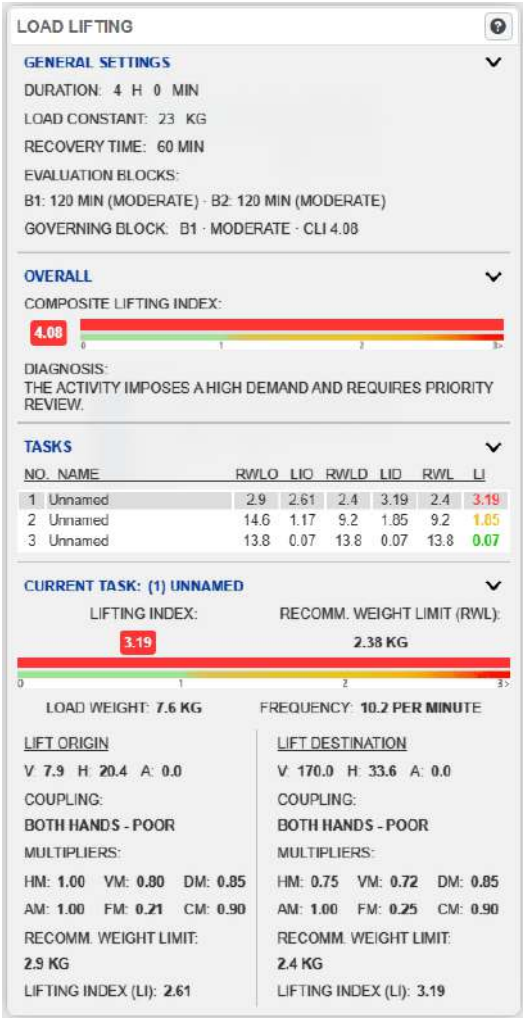


Figure 66: Main Load Lifting mode panel.

The general lifting settings are shared by all tasks in the analysis. They include temporal organization, the Load Constant, and anthropometric adjustment. Changing any of these parameters recalculates every task and every temporal block.

NIOSH tasks must be distinguished from temporal periods. A task describes a specific lifting condition. A period indicates how long the task profile is performed before recovery. The number of periods does not change the number of tasks, and the number of tasks does not determine the number of periods.

Ergoniza Intelligent uses a shared profile: all periods contain the same tasks at the same frequencies. Durations and

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recovery times can vary between periods, but a different set of tasks or frequencies cannot be assigned to each period.

Temporal organization can be entered in three ways (Figure 67):

- **One lifting period**: represents a continuous activity with a single duration and does not require a subsequent recovery time.
- **Regular pattern of periods**: represents several periods of equal duration separated by equal recovery times. Users enter the duration of each period, the number of periods, and the recovery time between periods.
- **Custom sequence of periods**: allows periods with different durations and recovery times to be defined and arranged. The recovery time entered after a period occurs before the next period; the last period does not require a subsequent recovery time.

Ergoniza Intelligent uses recovery times to determine whether periods are evaluated independently or must be combined into the same evaluation block. At each boundary between periods, the available recovery time, RT, is compared with the lifting time accumulated in the block, WT. If the recovery meets the applicable requirement, the block ends and the next period starts a new block. Otherwise, the recovery is considered insufficient to separate the periods, and their lifting times are accumulated.

When insufficient recovery requires periods to be combined, the duration category is determined from the lifting time accumulated in the resulting block. Once the long-duration category has been reached, the Revised NIOSH Lifting Equation assumes ordinary industrial rest allowances and does not establish an RT/WT ratio that would allow subsequent periods in the same block to be separated again.

For short-duration blocks, with up to 60 minutes of accumulated lifting time, recovery must be at least equal to the lifting time; that is, RT/WT ≥ 1.0. For moderate-duration blocks, with more than 60 and up to 120 minutes of lifting, RT/WT must be ≥ 0.3. Blocks lasting more than 120 and up to 480 minutes are classified as long duration and are evaluated assuming ordinary industrial rest allowances. NIOSH does not provide weight limits for more than eight hours of lifting.

The temporal summary shows total lifting time, total recovery time, total elapsed time, and the resulting evaluation blocks. Total elapsed time is the sum of lifting time and recovery times in the sequence.

All evaluable blocks are calculated and compared. The LI is compared in a single-task analysis, and the CLI is compared in a multitask analysis. The block with the most unfavorable index is identified as the governing block. If several blocks reach the same maximum value, all of them share this status. Indices from different blocks are neither added nor averaged.

Ergoniza Intelligent considers the temporal organization non-evaluable when a block accumulates more than eight hours of lifting or when the total elapsed time of the sequence, including recovery times, exceeds eight hours. In that case, no valid overall NIOSH result is presented and no governing block can be identified.

The **Load Constant, LC**, is the initial value to which the six multipliers are applied. The conventional NIOSH formulation uses **23 kg**. Ergoniza Intelligent allows values from 20 to 30 kg to be selected, but any value other than 23 kg modifies the standard method and should only be used when there is an explicit methodological justification.

When several tasks are present, the program calculates the RWL and LI for each one and also obtains a **Composite Lifting Index** for the task profile in each required duration category. This index is neither the mean nor the maximum of the individual indices: it incorporates the effect of the cumulative frequencies of the different tasks using the procedure described in Section 13.5.

The Revised NIOSH Lifting Equation uses standard geometric references: an optimal vertical height of 75 cm, a maximum height of 175 cm, and horizontal limits of 25 and 63 cm. These references do not vary with the dimensions of the person being assessed. When **anthropometric adjustment** is enabled, Ergoniza Intelligent replaces these references with values derived from the model's dimensions.

Specifically, the system adapts the vertical lifting reference, the maximum height considered, and the horizontal limits used in the calculation based on the worker's anthropometry.

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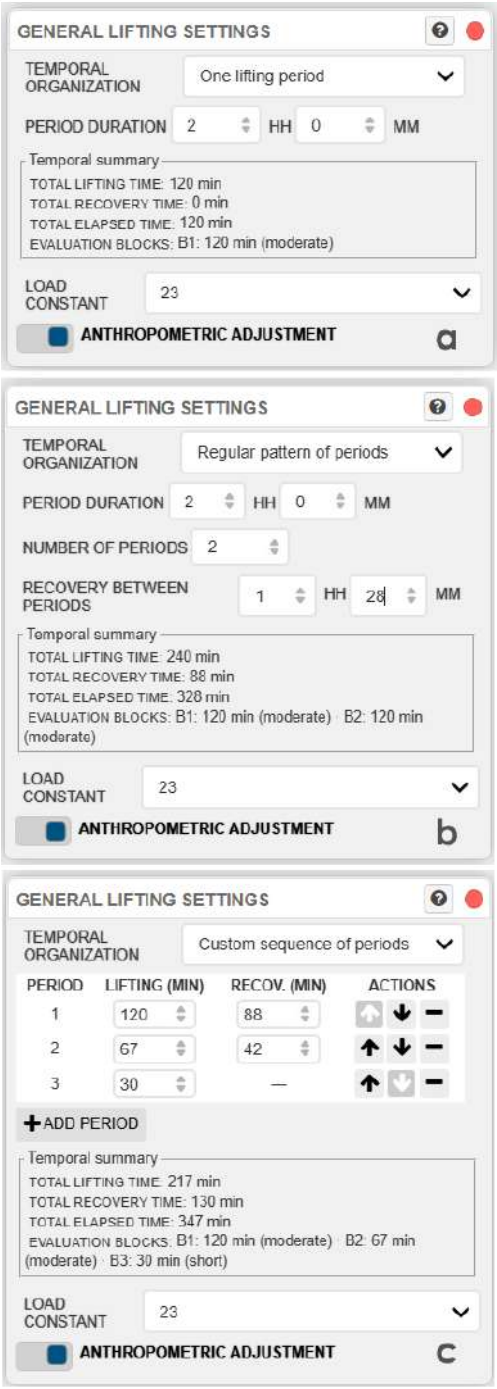


Figure 67: Temporal organization modes for lifting: (a) one period; (b) regular pattern of periods; (c) custom sequence of periods.

This change can affect several equation multipliers, particularly those related to the load's horizontal distance, vertical height, vertical travel distance, frequency, and coupling quality.

This option introduces an adaptation of the conventional application of the NIOSH equation[81]. Therefore, when anthropometric adjustment is enabled, the results should not be interpreted as a strictly standard application of the method, but as an estimate adjusted to the body dimensions of the person being analyzed. This option may be useful when assessing how a worker's anthropometric characteristics may influence actual lifting conditions.

13.3 Origin and destination positions

Each lifting task must be defined by two main positions: the load's **origin** and **destination**. The origin is the initial position from which the load is picked up or begins to be raised, whereas the destination is the final position where the load is placed or reaches the endpoint of the lift.

In Ergoniza Intelligent, these positions can be defined manually on the three-dimensional model or by capturing postures with artificial intelligence. They characterize the load's location relative to the worker (Figure 68).

The Revised NIOSH Lifting Equation recognizes that conditions at the origin and destination may differ. Ergoniza Intelligent calculates and displays the corresponding factors for both points. However, the destination can determine the task result only when significant control of the load is required at that point.

The **horizontal distance** expresses the load's distance from the worker's body. The **vertical height** indicates the load's position relative to the floor. Lifts performed from very low positions or to very high positions are usually less favorable than lifts performed near the method's reference height.

The **vertical travel distance** is the difference in height between the origin and destination. This value accounts for the lift's vertical range. **Asymmetry**, in turn, accounts for whether the lift involves rotation or deviation from the worker's frontal plane. Asymmetric lifts reduce the Recommended Weight Limit because they create less favorable conditions than a symmetrical lift performed in front of the body.

The system displays the results associated with the origin and destination to facilitate review of the movement. When **Significant Control of the Load Is Required at the Destination** is enabled, both endpoints are compared and the task adopts the lower RWL and the higher resulting LI. When it is disabled, the task result is determined solely from the origin;

[81] Moehamad Adi Rochmat et al., "Modifying the Revised NIOSH Lifting Equation Using Anthropometric Variables to Calculate Horizontal and Vertical Multipliers," in *2022 4th International Conference on Cybernetics and Intelligent System, ICORIS 2022*, 2022, 1–8, doi:10.1109/ICORIS56080.2022.10031509.

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destination results are retained as descriptive information but cannot govern the task index.

This setting is stored independently for each task. Therefore, in a multitask analysis, significant control at the destination may be required for some tasks but not for others.

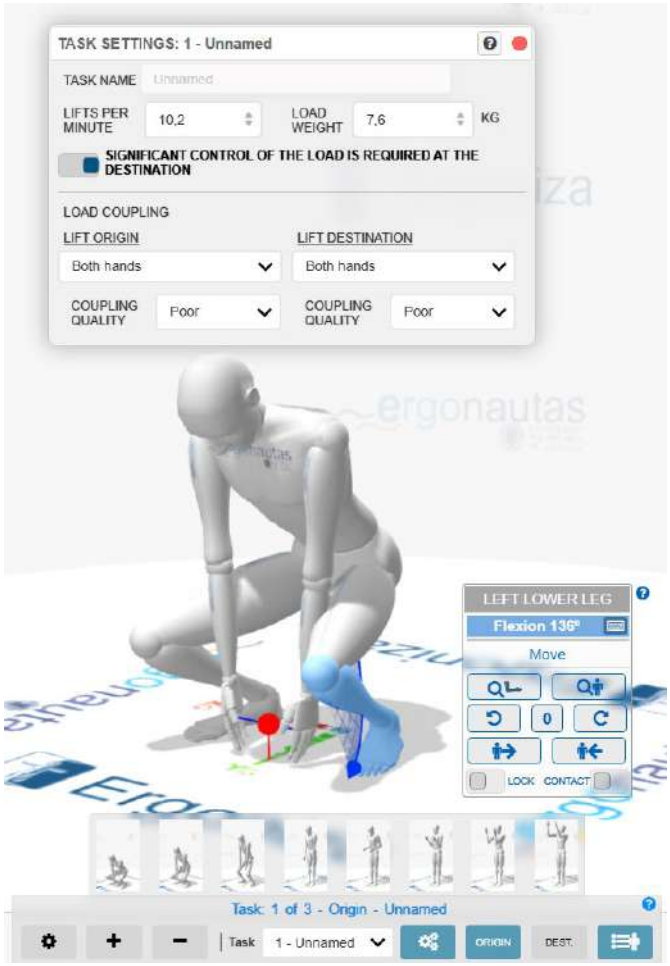


Figure 68: Defining lift origin and destination.

When the load position, the worker's posture, or any task parameter is changed, the results are updated automatically.

13.4 Weight, frequency, and coupling

In addition to the origin and destination positions, each lifting task must be characterized by variables describing the load handled and the exposure conditions. These variables are entered in the **Task Settings**, as shown in Figure 68.

The **load weight** is the actual weight handled by the worker. This value is subsequently compared with the Recommended Weight Limit calculated by the system. Therefore, the entered weight does not change the RWL; instead, it is used directly to calculate the Lifting Index by relating the actual load to the recommended limit for the task conditions.

The **frequency F** is the average number of lifts per minute for the task. When the frequency entered falls between two NIOSH tabulated values, Ergoniza Intelligent obtains the FM through linear interpolation. The FM is calculated for each duration category required by the evaluation blocks and takes the load's vertical height into account. The same task can therefore have different FM, RWL, and LI values depending on whether it is evaluated in a short-, moderate-, or long-duration block.

The **coupling quality** can be defined independently at the origin and destination as Good, Fair, or Poor. Together with the load's vertical height, this classification determines the Coupling Multiplier **CM** at each endpoint. A less favorable coupling reduces the RWL.

The **handling method** must also be specified, particularly whether the load is handled with one hand or both hands. This information is necessary to represent the analyzed situation accurately and to interpret the results in relation to the actual task.

13.5 Analysis results: Recommended Weight Limit, Lifting Index, and Composite Lifting Index

Once the task parameters have been defined, Ergoniza Intelligent calculates the main lifting-analysis results. These results are displayed in the **Load Lifting** panel, as shown in Figure 65.

A key result is the **Recommended Weight Limit**, or **RWL** (*Recommended Weight Limit*). This value is a reference limit for the specific task conditions and duration category being evaluated. It should not be interpreted as an individual maximum weight or a guarantee that injury will not occur. To obtain it, the system starts with the Load Constant and applies the multipliers defined by the Revised NIOSH Lifting Equation (Equation 56).

The RWL is used to calculate the **Lifting Index**, or **LI**, defined as the ratio of the actual load weight to the Recommended Weight Limit (Equation 55). A value of **LI ≤ 1** indicates that the actual weight does not exceed the calculated RWL. If **LI > 1**, the load exceeds the recommendation for the conditions considered and the task should be reviewed. The index should be interpreted as a relative indicator of physical demand, not as a prediction of injury for an individual.

The index must be interpreted by considering all task conditions. A high LI may be due to the load weight, but it may also result from excessive horizontal distance, an unfavorable origin or destination height, high frequency, prolonged duration,

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marked asymmetry, or poor coupling quality. The multipliers should therefore also be reviewed to identify which conditions are penalizing the lift.

When the analysis contains several tasks, the system also calculates the **Composite Lifting Index**, abbreviated **CLI**, defined by the NIOSH methodology. To obtain it, tasks are ordered from highest to lowest individual index. The most demanding task contributes its full LI, and the remaining tasks contribute the incremental demand produced as their frequencies are progressively added to the cumulative frequency.

The complete task profile is evaluated for every duration category present in the temporal blocks. The same task can therefore have different FM, RWL, and LI values when evaluated as part of a short-, moderate-, or long-duration block. The CLI can also change in a multitask analysis.

All evaluation blocks are compared. In a single-task analysis, the governing block is the one with the most unfavorable LI. In a multitask analysis, it is the one with the most unfavorable CLI. If several blocks reach the same maximum value, all of them are governing blocks. The overall assessment result is the index of the governing block. Results from the other blocks provide context about the activity's temporal progression, but they are neither added to nor averaged with the governing result.

When a block accumulates more than eight hours of lifting or the total elapsed time of the sequence exceeds eight hours, the temporal organization is considered non-evaluable. In that case, Ergoniza Intelligent does not present a valid overall LI or CLI and does not identify a governing block.

Ergoniza Intelligent allows these results to be used not only to assess the existing task but also to explore redesign alternatives. By changing the load position, reducing horizontal distance, improving coupling, decreasing frequency, limiting duration, or adjusting the weight handled, the evaluator can see how the RWL and Lifting Index change.

14 POSTURE SEQUENCE ANALYSIS

Many work activities cannot be adequately represented by a single posture. In these cases, the body's evolution over time must be described by recording several successive postures that represent a task, a work cycle, or a significant part of an activity.

Ergoniza Intelligent supports **posture sequences** in *Multiple Postures* mode. In this mode, the human model does not represent only one body configuration at a single point in time, but an editable animation made up of postures placed at different points on a timeline. Instantaneous and time-dependent analyses can be applied to this sequence, including cumulative exertion load, changes in postural discomfort, and the assessment of postures sustained during an activity.

Ergoniza Intelligent can create posture sequences using several procedures. The most appropriate method depends on the type of task, the available information, and the required degree of accuracy.

The most direct way to create an animation is to build it manually from the graphical model. The user can place the model in an initial posture, record it, move forward in time, change the posture, and record it again, repeating the process as many times as necessary.

Manual creation provides maximum control over the sequence because the user decides exactly which postures are included, the order in which they appear, and the duration of each interval between them.

When one or more representative images of the task are available, they can be used to generate postures that are then incorporated into an animation. A posture can be entered manually from an image or captured using the artificial-intelligence posture-detection system (Section 14.3). This procedure is particularly useful when the task does not require a complete continuous reconstruction, but rather a representation of key phases, such as the beginning, an intermediate phase, and the end. Several images can be used to build a temporal sequence that is sufficiently representative for analysis.

Animations can also be generated in Ergoniza Intelligent by capturing multiple postures from a video of the activity. The system can load a video from the user's device, select the relevant time segment, and automatically record postures throughout the sequence using artificial intelligence. The video-capture interface allows users to choose between Fast AI and Accurate AI (Section 14.3), and, for Fast AI, set the sampling rate at which video postures will be recorded.

This procedure is appropriate for analyzing an actual activity performed by a person under natural working conditions. The user can select the video interval to be studied and automatically generate a posture sequence on the timeline.

When data from motion-capture systems are available, Ergoniza Intelligent can import them and convert them into an animation of the biomechanical model. In particular, the system supports the import of Xsens MVN data in MVNX format. This option is appropriate when a detailed temporal description of movement is required, especially in instrumented studies or complex tasks. Section 14.4

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describes the import of data files from motion-capture (MOCAP) systems in greater detail.

14.1 Manually creating a posture sequence

A sequence can be created manually by changing the model's posture and successively recording the body positions relevant to the activity. The user can edit the posture with the controls available for each body segment and, once the desired position has been defined, record it at the current point on the timeline.

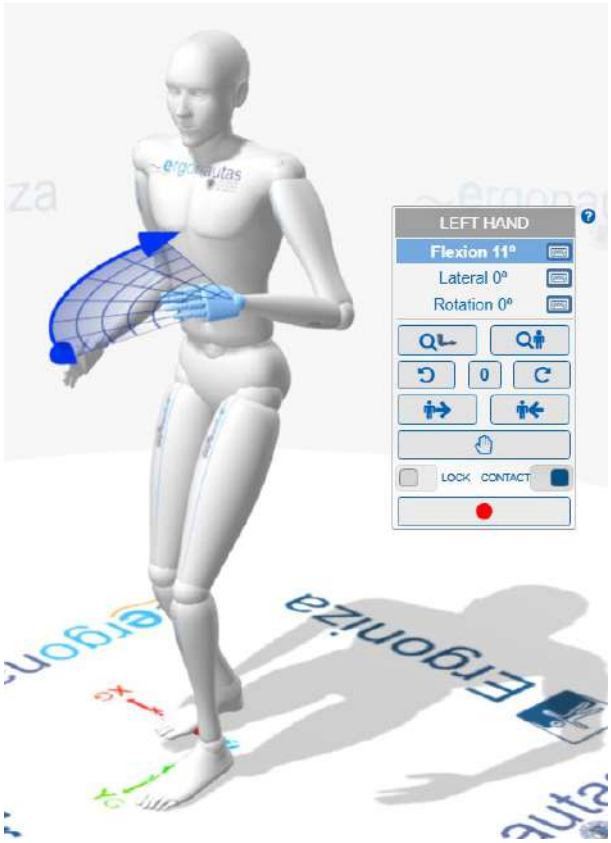


Figure 69: Recording postures in animation sequences.

A posture is recorded using the posture record button, identified by a red dot in the editing panel, as shown in Figure 69. Pressing this button stores the model's current posture at the selected animation time.

Each recorded posture appears on the timeline as a small rectangle representing a **keyframe** (Figure 70). These keyframes indicate the points in time at which the user has explicitly defined a posture. The animation is built from the succession of these keyframes and the duration separating them.

To create an animation manually, the general procedure is to move to a point on the timeline, change the model's posture, record the posture, move to another point, change the model again, and record a new posture. By repeating this process, the user can build a sequence representing the main phases of the activity.

Keyframes created manually by the user, except for the initial one, can be moved along the timeline to adjust when each posture occurs. Keyframes obtained through automatic capture retain their sampling times and cannot be moved individually from the timeline. Both types can be deleted when they are no longer needed or when the animation's temporal structure must be changed. The sequence is therefore not irreversibly fixed when it is created; it can be edited later to improve its correspondence with the actual task.

14.2 Timeline, interpolation, and animation editing

The timeline controls the evolution of the posture sequence. It shows the temporal position of each keyframe, the current playback time, and the main animation-editing controls, as shown in Figure 70.

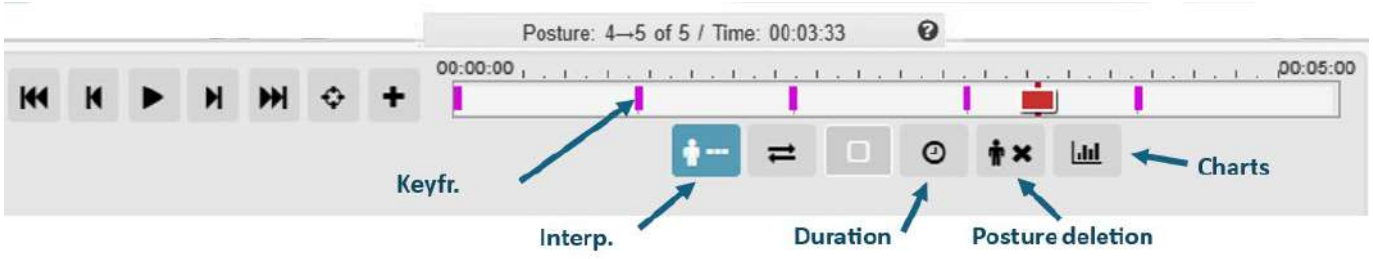


Figure 70: Animation timeline.

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When interpolation is enabled, Ergoniza Intelligent generates intermediate postures in segments whose transitions are defined as interpolable, especially between keyframes created or edited by the user. Segments made up exclusively of automatically captured postures—for example, from a video or an MVNX file—retain the recorded sample sequence and are not automatically interpolated again. Interpolation represents a gradual transition without requiring every point in the movement to be defined manually.

When interpolation is disabled, transitions between postures are not represented progressively. Instead, each recorded posture is maintained until the next keyframe is reached. This option can be useful for representing discrete posture changes or when the system should not generate intermediate positions.

Timeline controls also allow users to select or deselect the entire sequence, change the animation's total duration, delete selected postures, and access graphical representations of the analyzed variables. These functions facilitate the editing of existing sequences and make it possible to adjust the relative duration of the activity's different phases.

In a posture sequence, explicitly recorded postures must be distinguished from postures interpolated by the system. Recorded postures are those created or captured by the user and displayed as keyframes on the timeline. Interpolated postures are intermediate positions automatically generated in segments whose transitions are defined as interpolable.

This distinction is important when interpreting results. A recorded posture represents a body configuration defined by the user or captured from an external source. An interpolated posture, in contrast, is an intermediate estimate calculated by the system to complete the animation's temporal evolution. Therefore, when accuracy during a particular phase is especially important, additional keyframes should be recorded instead of relying exclusively on broad interpolation between two very different postures.

The quality of a manual sequence depends on selecting the appropriate keyframes. If too few postures are recorded, the animation may not adequately represent relevant changes in movement. If too many are recorded, editing may become more complex without necessarily providing a significant improvement in ergonomic interpretation. As a general rule, users should record postures that define relevant changes in the position of the trunk, arms, legs, or head, in the loads handled, or in support conditions.

14.3 Artificial intelligence for posture detection

Automatic posture detection using **artificial intelligence** in Ergoniza Intelligent relies on computer-vision and machine-learning techniques that estimate anatomical landmarks and the relative orientation of body segments from images and video sequences. From these estimates, Ergoniza Intelligent reconstructs a three-dimensional posture of the model and calculates the corresponding joint angles. Tracking and smoothing mechanisms are applied to video to reduce noise between frames. The results are estimates and must be reviewed by the user, especially when there are occlusions, unfavorable viewpoints, blur, or rapid movements.

14.3.1 Two artificial-intelligence modes in Ergoniza Intelligent

Ergoniza Intelligent implements posture detection using two complementary approaches that address different privacy, speed, and accuracy requirements. The first, called **Fast AI**, performs inference on the user's device using code that runs in the browser. This approach considerably reduces response latency and, most importantly, avoids sending images and videos to Ergonautas servers, preserving the privacy of the audiovisual material at its source.

The second, called **Accurate AI**, delegates processing to Ergonautas server infrastructure. This option prioritizes estimation quality and robustness in complex scenarios, in exchange for longer processing time and the need to transfer data outside the local device. The user experience is consistent in both cases: the user provides the capture, reviews the result projected in the Ergoniza Intelligent interface, makes any appropriate adjustments, and immediately applies the detected posture data to the available assessment methods.

14.3.1.1 Fast AI (on the user's device)

In **Fast AI** mode, all detection is performed locally and media files never leave the user's device. This technical choice provides quick responses that facilitate iteration during capture—for example, repeating a shot or refining the framing—and prevents the selected image or video from being sent to Ergonautas servers for inference.

This mode supports both images and video sequences. For images, estimation focuses on a single posture representative of the captured moment. It immediately overlays anatomical landmarks and segments on the photograph and makes the resulting joint angles available in the interface. This facilitates a

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quick visual check and, if necessary, a specific correction by the evaluator.

For video, the system processes the complete sequence and maintains sufficient basic consistency between frames to observe the temporal evolution of a task, identify posture changes, and extract the metrics required for ergonomic analysis.

Practical performance depends on the device's capabilities—including its CPU and the acceleration available in the browser—the quality of the capture, and the absence of substantial occlusions. When lighting is uniform, the full body is visible in the frame, and a three-quarter view is used, local detection can provide suitable results for an initial estimate. The evaluator must review that estimate before making decisions, especially when speed and privacy are priorities.

14.3.1.2 Accurate AI (on Ergonautas servers)

In **Accurate AI** mode, inference is transferred to Ergonautas servers, where specialized pipelines and computing resources can improve estimation quality under demanding conditions. In this approach, the user authorizes the image or video to be sent to the platform, which returns a more stable result and a more robust interpretation of complex scenes involving occlusions, less favorable viewpoints, or movements with extreme ranges.

For images, the system reconstructs the posture from a single photograph. This is especially useful when no video is available or temporal continuity is not required, and it provides broad anatomical coverage suitable for technical documentation.

For video, the platform generates a coherent time series of postures and angles, stabilized against spurious variations between frames. This facilitates the identification of work-cycle phases, exertion peaks, and limiting postures. Results are delivered in structured formats for direct integration into Ergoniza Intelligent's ergonomic assessment methods and for traceability in reports. The improved accuracy and robustness come at the cost of greater latency associated with data transfer and remote processing, which is generally acceptable when the primary objective is to obtain a higher degree of confidence in the estimate.

14.3.1.3 Image and video capture modes

In both Fast AI and Accurate AI modes, Ergoniza Intelligent can work with still images and video, adapting to the ergonomic problem being addressed. With images, the tool is designed to assess individual postures from representative photographs. The professional can clearly review the estimated overlay on the capture, confirm that key points are correct, and make adjustments if any misalignment due to occlusion or an unfavorable viewpoint is detected. This approach is particularly efficient for preparing technical records, documenting individual cases, and extracting static metrics without requiring temporal continuity. With video, the tool analyzes dynamic tasks and generates a posture sequence over time, making it possible to study kinematic evolution, detect critical moments, and compare the effects of changes in work methods or workstation design. The temporal consistency obtained in this mode facilitates the interpretation of movement patterns and provides richer context for risk assessment.

14.3.1.4 Workflow and selection criteria

The workflow is designed to be simple and consistent. First, the user selects the processing approach that best matches the priorities of the case: speed and privacy with local processing, or enhanced accuracy with server processing. The user then decides whether the case requires an image or video capture. After providing the material, the user views the estimated overlay returned by the system, confirms or corrects the relevant details, and immediately applies the posture data to the available assessment methods. The choice between the two approaches generally depends on the balance among response time, data sensitivity, and case complexity. Fast AI is appropriate when confidentiality and immediate results are the priorities. Accurate AI can be selected when remote processing is preferred and greater latency and transfer of the audiovisual material are acceptable. In both cases, the resulting posture must be reviewed and validated.

14.3.1.5 Capture best practices

Result quality depends substantially on how the capture is made. It is advisable to frame the person's entire body from head to toe, maintain uniform lighting that avoids backlighting and harsh shadows, and choose a viewpoint that reduces overlap and makes the relevant joints visible; a three-quarter view may be appropriate for many tasks. For video, the camera should be kept stable and the frame rate should be sufficient to describe the movement clearly. Following these guidelines increases estimation reliability in both Fast AI and Accurate AI modes and reduces the need for subsequent manual corrections.

14.3.2 AI versus MOCAP: scope, limitations, and user validation

Motion-capture (MOCAP) systems, including marker-based systems and certain laboratory markerless systems, are a reference when highly accurate and repeatable kinematic measurement is required. These systems operate with

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calibrated cameras and controlled geometric models of the environment, enabling posture reconstruction with very low uncertainty and explicit control of scale, relative sensor positions, and temporal synchronization. In contrast, artificial-intelligence posture detection from general-purpose images or video relies on algorithms that must "infer" body structure from uncalibrated 2D signals, often under suboptimal lighting, occlusion, and perspective conditions. For this reason, although AI provides a fast, accessible, and potentially useful solution for numerous ergonomic analyses when the result is reviewed, its accuracy does not reach that of a properly configured MOCAP system.

The main sources of discrepancy between the two approaches arise from the nature of the problem that AI must solve. First, there is inherent 2D-to-3D ambiguity: different three-dimensional body configurations can project very similar images when the camera is not calibrated or the scene contains no metric references. Second, scale and viewpoint introduce systematic biases if the subject is not appropriately framed or the camera is sharply tilted; AI compensates using anthropometric assumptions and statistical priors that do not always match individual morphology. Third, partial occlusions caused by clothing, tools, furniture, or overlapping segments reduce the visibility of certain joints and require the model to estimate unseen positions. This can result in displaced joints, implausible extreme angles, or local instability. Finally, in video sequences captured in real-world environments, the effective frame rate, motion blur, and variations in lighting increase temporal noise and may require subsequent smoothing, which can attenuate very brief peaks.

With these considerations in mind, Ergoniza Intelligent incorporates automatic posture detection as a rapid capture and preliminary-analysis mechanism fully integrated with the software's assessment methods. However, it follows a clear operating principle: when a posture originates from AI, the user must review and validate the result before incorporating it into the analysis. The interface facilitates this verification by overlaying the estimate on the image or video and displaying the resulting joint angles. If an inconsistency is identified because of occlusion, poor interpretation of a segment, or inappropriate framing, the evaluator can correct the required elements and confirm the result again. This validation is not a bureaucratic formality; it verifies whether the data used in the assessment adequately represent the observed posture and whether capture limitations have been taken into account.

In terms of recommended use, posture AI may be suitable for assessing common static or dynamic tasks, conducting rapid screenings, comparing scenarios before and after an intervention, and documenting representative situations without instrumented testing. Conversely, when the study requires reference-level accuracy—for example, biomechanical research validation, detailed monitoring of micromovements, synchronization with force or EMG data, or certification where the allowable error is very low—the appropriate resource is a MOCAP system with a corresponding calibration and quality-control protocol.

Good practice in Ergoniza Intelligent therefore consists of capturing with careful attention to framing, lighting, and viewpoint; reviewing the overlay of anatomical landmarks and segments; confirming that key angles are plausible for the task being analyzed; and making the necessary adjustments whenever there is any indication of occlusion or questionable interpretation.

14.4 Importing data from Xsens

Ergoniza Intelligent can import movement data acquired with inertial motion-capture systems such as Xsens MVN. Xsens MVN records human movement in three dimensions and is used in fields including ergonomics, biomechanical research, sports medicine, film, video games, and animation. It uses a network of inertial sensors placed on the body that contain accelerometers, gyroscopes, and magnetometers to measure linear acceleration, angular velocity, and the magnetic field, respectively. The system can capture movement in real time and transmit the data to a computer for subsequent analysis or processing.

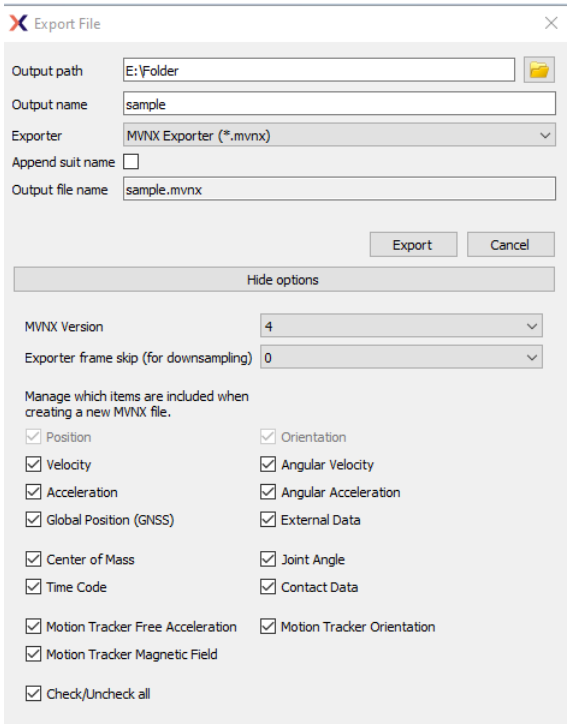


Figure 71: Xsens MVN dialog box with options for exporting to MVNX format

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Xsens MVN systems can export acquired user-movement data in MVNX (MVN Open XML) format, which can be imported by Ergoniza Intelligent. MVNX files from version 3 through the version current at the date of this document can be imported. Figure 71 shows the Xsens MVN dialog box with options for exporting to MVNX format. Of the dynamic data options shown in Figure 71, Ergoniza Intelligent requires Position and Orientation data. The importer also reads the segment definitions and required timing information, including the time of each frame and the sampling rate.

After the file to be imported is selected, Ergoniza Intelligent reads its original sampling rate. When the file contains more than 100 frames and its rate is higher than 30 frames per second, the program prompts the user to reduce it to 15, 20, or 30 fps (Figure 72). The user can choose a value below 30 fps if the device running Ergoniza Intelligent does not have sufficient capacity to display postures at an adequate speed. When selecting this value, it should be considered that the sampling rate can affect the accuracy with which body-segment accelerations are estimated. Sampling rates below 12 frames per second may underestimate peak accelerations by 22%, whereas rates above 100 provide little improvement in estimating the acceleration profile[82].

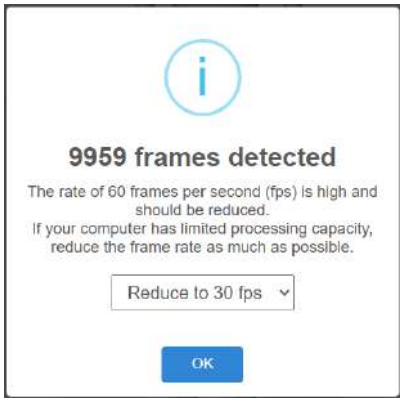


Figure 72: MVNX file import options dialog box.

From the positions and orientations of the different body segments in MVNX files, Ergoniza Intelligent reconstructs the model's postures over time and calculates the user's estimated stature and the lengths of the different body segments. The user's weight is not imported and must therefore be entered manually.

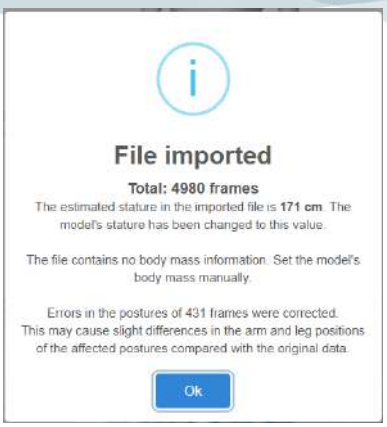


Figure 73: MVNX file import summary dialog box.

The import result is presented to the Ergoniza Intelligent user in a dialog box such as the one shown in Figure 73.

14.4.1 Adapting MVNX data to the Ergoniza Intelligent model

As explained in Section 5.1.4, the forearm and lower-leg segments of the Ergoniza Intelligent model have one degree of freedom (flexion/extension). Radioulnar rotation and tibiofibular rotation are delegated to the next segment in the kinematic chain—the hand for the forearm and the foot for the lower leg. Forearm flexion relative to the upper arm must therefore occur in the plane formed by the shoulder, elbow, and wrist joints, and lower-leg flexion relative to the thigh must occur in the plane formed by the hip, knee, and ankle joints.

This is illustrated in Figure 74, which shows the forearm rotating about the elbow in a flexion movement. If there is no humeral rotation at the shoulder, all wrist positions over time (Mt_1, Mt_2, Mt_3...) and the elbow and shoulder joints remain in the same plane.

[82] S. Kromodihardjo and A. Mital, "The Cost/Accuracy Approach: A Method for Selecting Filming Rate for Manual Materials Handling Activities," *Applied Ergonomics* 17, no. 3 (1986): 218–20, doi:10.1016/0003-6870(86)90011-6.

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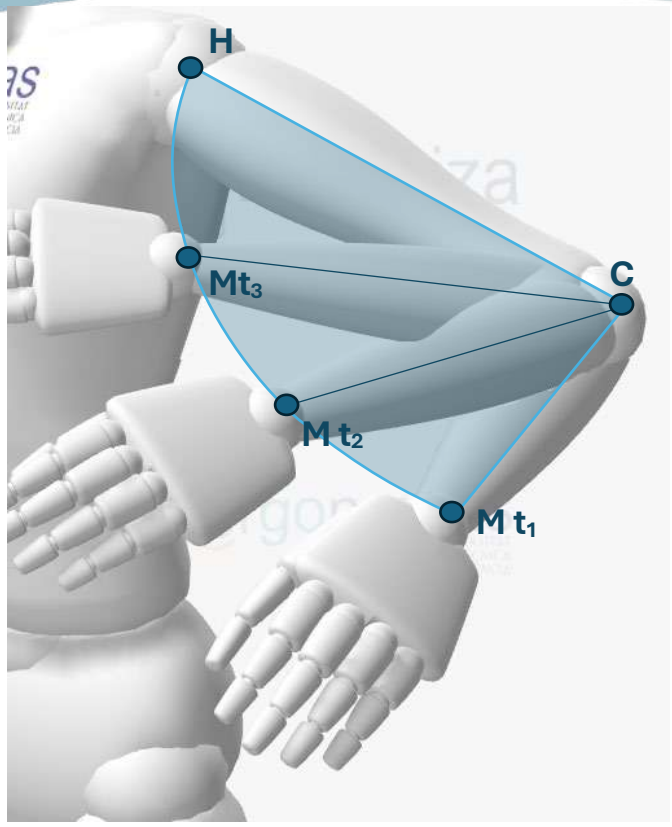


Figure 74: Plane of forearm flexion about the elbow.

When data are imported from MVNX files, the axial orientations of the upper arm and forearm—that is, their rotations about the longitudinal axis—are adapted to this requirement. The plane formed by the shoulder, elbow, and wrist is calculated, and the longitudinal axes of the upper arm and forearm are oriented so that the direction of forearm flexion lies in that plane. A similar procedure is applied to the thigh, lower leg, and foot.

When the forearm or lower leg is fully extended, the plane containing the corresponding joints cannot be determined because the joints lie on the same line. Therefore, when the forearm or lower leg is fully extended, the axial orientation of the upper arm is adopted for the forearm.

Finally, the inherent inaccuracy of inertial sensors can cause forearm or lower-leg extensions slightly greater than 180º to be recorded. In these cases, Ergoniza Intelligent corrects the forearm or lower-leg extension to 180º and notifies the user of the correction, as shown in Figure 73.

14.5 Loads and conditions that vary throughout a posture sequence

In Multiple Postures mode, the temporal sequence represents not only changes in the model's posture but also variations in certain analysis conditions over time. One of the most important of these is the **external load** applied during the activity.

Ergoniza Intelligent allows a load to be defined for a specific posture or a time interval in the sequence. This can be done by selecting part of the posture sequence on the timeline, right-clicking, and choosing **Define Load**, or by pressing the **Load** button in the left sidebar.

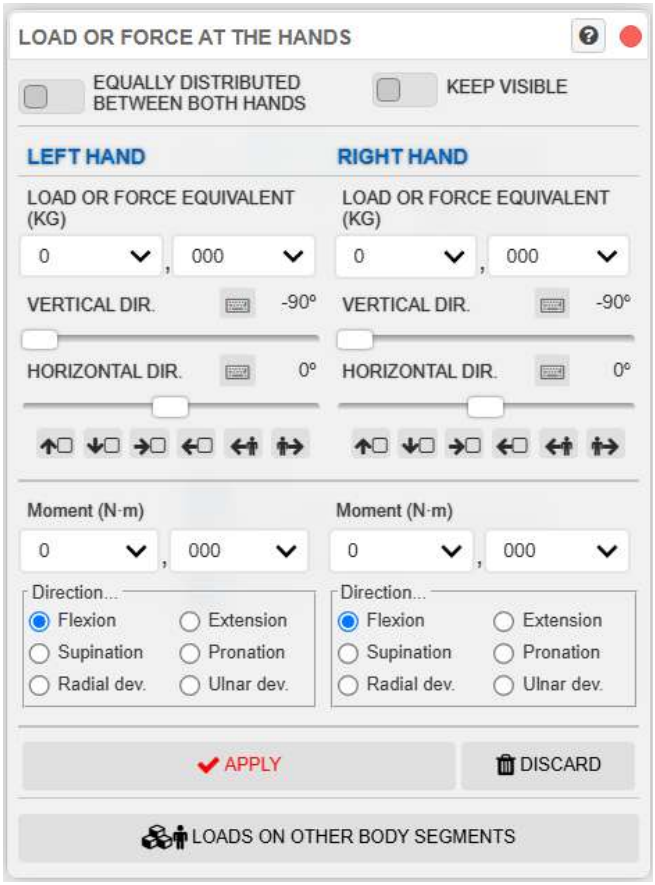


Figure 75: Load or force at the hands dialog box.

This opens the dialog box shown in Figure 75, where the load, force, or moment at the hands can be defined for the selected portion of the posture sequence. The analysis is therefore not limited to a single constant load throughout the animation; it can represent situations in which the applied or handled load changes over the course of the task.

This feature is especially useful when an activity includes phases with different conditions. For example, the first part of the sequence may involve no load, followed by a phase in which an object is handled, and a final phase in which the load is no longer present. The magnitude or manner in which the load is applied may also vary at different points in the activity. In these cases, the system can explicitly represent this variability.

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When a load is applied to a posture or interval, that information becomes part of the analysis conditions for that time segment. The biomechanical and ergonomic results obtained for the sequence therefore depend not only on the adopted posture but also on the load defined at each point. This is particularly relevant when calculating biomechanical exertion, cumulative exertion load, lumbar compression, and other analyses sensitive to applied external forces.

In addition to the load, some specific analysis settings can be changed throughout the sequence according to the nature of the task. For example, certain methods may require particular posture or activity conditions to be adjusted for a specific time segment. The ability to apply settings to an interval makes it possible to adapt the sequence more accurately to the observed real-world situation.

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15 APPENDICES

15.1 APPENDIX 1: CALCULATION OF UPPER-ARM AND THIGH ROTATIONS

Calculating humeral and femoral rotation is conceptually more complex than calculating other joint angles because it is defined not only by the direction of the segment, but also by its orientation "around" its own longitudinal axis.

Focusing on humeral rotation as an example, without loss of generality, rotation is commonly estimated by defining a reference plane associated with the trunk and upper arm and comparing it with the functional plane of the upper arm and forearm. This approach is robust in most postures, but it has an inherent problem: certain configurations cause the reference plane to become poorly defined or extremely sensitive to small postural variations.

The typical case occurs when the upper arm is almost aligned with the trunk axis—for example, when it is nearly vertical alongside the body. In this situation, the humeral axis is almost parallel to the trunk axis, and the "trunk–upper arm" plane loses geometric stability: small abductions or minimal deviations can produce large changes in the plane's orientation and therefore abrupt jumps in the estimated humeral rotation. This phenomenon is an example of a kinematic singularity, equivalent to what is known as *gimbal lock* in Euler-angle representation systems: one effective degree of freedom is lost, and the angular decomposition is no longer unique or becomes numerically unstable.

There is also another degenerate case: when the shoulder, elbow, and wrist are almost aligned. Under this condition, the plane defined by these three points—which is used to describe the orientation of the upper-arm–forearm assembly—also becomes indeterminate, making it impossible to calculate a reliable rotation angle.

15.1.1 How Ergoniza Intelligent calculates humeral and femoral rotation

Once again, the explanation focuses on humeral rotation; the method is equivalent for the femur. The method geometrically defines humeral rotation as a dihedral angle—a relative rotation—about the humerus's longitudinal axis:

1. **Axis of rotation (humeral axis).** The upper arm's longitudinal axis is defined by the line connecting the shoulder and elbow. Humeral rotation is interpreted as rotation about this axis.
2. **Trunk-based reference plane.** A reference plane is constructed to represent the "base" orientation of the upper arm relative to the trunk. Conceptually, it is the plane formed by the humeral axis and the trunk's longitudinal axis. This plane provides the relative "zero" for measuring rotation about the humerus, and its reference is fixed to the trunk—that is, it follows the trunk's orientation.
3. **Upper-arm–forearm plane.** A second plane representing the instantaneous orientation of the upper-arm–forearm assembly is constructed from the shoulder, elbow, and wrist points. This plane captures how the forearm is oriented around the humeral axis.
4. **Dihedral-angle calculation.** Humeral rotation is obtained as the signed angle between the two planes, measured about the humeral axis. This angle distinguishes, for example, a position in which the elbow points forward from one in which it points outward, even when the upper arm has the same elevation.

This approach is consistent with a common biomechanical interpretation: humeral rotation appears as a change in the forearm's orientation around the upper-arm axis once a trunk-based reference has been established.

15.1.2 How singularities are resolved

As explained above, the plane-based calculation becomes unstable when the upper arm approaches alignment with the trunk (a *gimbal-lock* singularity). A regularization strategy is applied to ensure continuity and prevent numerical jumps:

- **Detection of proximity to the singularity**. The degree of alignment between the humeral axis and the trunk axis is evaluated. The more parallel—or antiparallel—they are, the greater the likelihood that the method based on the trunk–upper arm plane will become unstable.
- **Use of an alternative estimate of upper-arm "twist."** Near the singularity, rather than relying exclusively on the dihedral angle between planes—which can change abruptly in response to minimal variations—an alternative measure of rotation about the upper-arm axis is used because it is inherently more stable in this configuration. Conceptually, this alternative measure is based on the segment's own orientation relative to a trunk-fixed reference and behaves more reliably when the trunk–upper arm plane becomes ambiguous.
- **Progressive interpolation between the two estimates.** The final humeral-rotation value is not replaced abruptly; instead, it is obtained by blending the two estimates according to proximity to the singularity:

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1. **far from the singularity**, the dihedral angle between planes dominates because it provides more biomechanical information;
2. **near the singularity**, the alternative measure dominates because it is more stable;
3. **in the intermediate region**, a smooth, continuous transition is applied to prevent discontinuities.

- **Collinear shoulder–elbow–wrist case**. If the shoulder–elbow–wrist plane is degenerate because the points are collinear, the system cannot obtain a reliable dihedral angle and directly uses the alternative estimate, which does not depend on this plane.

In summary, humeral rotation is calculated using a geometric formulation based on the trunk–upper arm reference and the upper-arm–forearm plane. It is stabilized by a controlled transition to an alternative measure of rotation about the upper-arm axis when the joint configuration approaches conditions under which the plane-based definition loses uniqueness or becomes numerically fragile.

15.2 APPENDIX 2: Details of the calculation of postural discomfort in instantaneous or non-sustained postures

This appendix presents the mathematical procedure used by Ergoniza Intelligent to calculate postural discomfort in instantaneous or non-sustained postures, summarized in Section 11.2.3.3. Unlike the procedures applied to sustained or repetitive postures, this calculation does not estimate how discomfort evolves over time. Instead, it assesses the instantaneous proximity of each joint to the extreme regions of its effective range of motion.

This appendix provides the indicator's detailed formulation, including the smooth transition function used to prevent abrupt changes in the assessment, the method for determining the threshold at which instantaneous discomfort begins, the use of LUBA as a relative reference for the upper limbs, neck, and trunk, and the alternative procedure based on isocomfort equations for the hip, knee, and ankle. It also describes the operational age adjustment of effective ROM, used solely as a reference for calculating relative proximity to the joint limit.

The procedure presented here should be interpreted as a preventive ergonomic parameterization.

15.2.1 Transition function for instantaneous discomfort

As explained in Section 11.2.3.3, the calculation of instantaneous postural discomfort starts from the relative position of each joint within its effective range of motion. For each joint *j* and degree of freedom *d*, the value $r_{j,d}$, which expresses how close the adopted angle is to the effective ROM limit, is first obtained. This value is then compared with two references: the threshold $r0_{j,d}$, which marks the end of the central functional zone, and *r100*, which defines entry into the extreme zone of the effective ROM.

The purpose of the transition function is to convert this relative proximity into a percentage of instantaneous discomfort. While $r_{j,d}$ does not exceed $r0_{j,d}$, discomfort is considered zero. When $r_{j,d}$ lies between $r0_{j,d}$ and *r100*, discomfort progressively increases until it reaches 100%. If $r_{j,d}$ exceeds *r100*, the indicator can continue increasing up to a bounded maximum, set at 125%, indicating that the joint is moving even closer to the effective ROM limit.

To prevent abrupt changes when the angle exceeds the threshold of the comfortable zone, Ergoniza Intelligent uses a smooth transition function. First, a normalized variable *u, which can take values from 0 to 1*, is calculated:

$$u_{j,d} = \min\left[\max\left(\frac{r_{j,d} - r0_{j,d}}{1 - r0_{j,d}}, 0\right), 1\right]$$

When

$$r0_{j,d} < r_{j,d} < r100$$

Equation 57

Instantaneous discomfort is then calculated using a smooth transition function:

$$\mathrm{D_{puntual,j,d}} = 100 \cdot \left(3\mathrm{u_{j,d}^2} - 2\mathrm{u_{j,d}^3}\right)$$

Equation 58

This function satisfies three conditions:

$$\mathrm{D_{puntual,j,d}} = 0 \text{ when } r_{j,d} \le r0_{j,d}$$

$$\mathrm{D_{puntual,j,d}} = 100 \text{ when } r_{j,d} = r100$$

and the increase between the two values occurs gradually, without abrupt discontinuities.

When the joint exceeds r100 and continues approaching the effective ROM limit, Ergoniza Intelligent allows the indicator to exceed 100%. A second smooth transition between r100 and 1 is calculated:

$$u_{exc,j,d} = \min\left[\max\left(\frac{r_{j,d} - r100}{1 - r100}, 0\right), 1\right]$$

Equation 59

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Instantaneous discomfort is again calculated using a smooth transition function:

$$D_{puntual,j,d} = 100 + (D_{max} - 100)\left(3u_{exc,j,d}^2 - 2u_{exc,j,d}^3\right)$$

Equation 60

where *Dmax* is the maximum indicator value when the effective ROM limit is reached. In the current implementation, this value is 125.

Values above 100% should not be interpreted as an exact psychophysical measure of discomfort, but as an indication that the joint has exceeded the operational threshold for an extreme posture and is moving even closer to the effective ROM limit.

15.2.2 Determining r0 for the upper limbs, neck, and trunk

The value $r0_{j,d}$ is not the same for every joint. Available evidence shows that postural tolerance depends on the body region and movement being analyzed. Therefore, Ergoniza Intelligent uses different values for each joint or functional group. For the wrist, elbow, shoulder, neck, and trunk, the LUBA method[83] is used as a source of relative sensitivity. LUBA assigns discomfort scores to different angular ranges of upper-limb, neck, and trunk movements. These scores are not used as the direct result for instantaneous discomfort, but as a criterion for determining when the indicator should begin to increase.

For each joint *j* and degree of freedom *d*, two values are extracted from the LUBA tables:

- $S_{max,j,d}$: maximum score assigned by LUBA to that movement;
- $\theta_{entry,j,d}$: angle at which the movement enters the range with the maximum LUBA score.

The maximum score represents the relative severity of the movement. Movements with higher maximum scores are considered more sensitive and should therefore begin to be penalized earlier. A severity ratio is calculated for this purpose:

$$p_{j,d} = \frac{Smax_{j,d} - Smin}{Smax_{global} - Smin}$$

Equation 61

where *Smin* = 5 and $Smax_{global}$ = 15 are the minimum and maximum values used to normalize LUBA scores.

An initial *r0* value is obtained from this relative severity:

$$r0_{score\ j,d} = r0_{max} - (r0_{max} - r0_{min}) \cdot p_{j,d}$$

Equation 62

where $r0_{min}$ = 0.60 and $r0_{max}$ = 0.85.

Thus, a movement with low relative severity begins to be penalized closer to the end of the ROM, whereas a movement with high severity begins to be penalized earlier.

However, the maximum score alone is insufficient. Two movements can have the same maximum score but reach it at very different angles. In LUBA, for example, certain shoulder-rotation movements may reach their maximum-score range at much smaller angles than other movements with the same maximum score. Therefore, Ergoniza Intelligent introduces a second criterion based on the angle at which the maximum-score range begins.

First, the following is calculated:

$$r_{entry\ j,d} = \frac{\theta_{entry\ j,d}}{ROMef_{,j,d}}$$

Equation 63

Next, the relative level that the curve should have when this entry angle is reached is defined. This level depends on the movement's relative severity:

$$u_{entry,j,d} = u_{min} - (u_{max} - u_{min}) \cdot p_{j,d}$$

Equation 64

where u_{min} = 0.20 and u_{max} = 0.75.

This value is used to solve for the *r0* that makes the transition curve reach that level at *rentry*:

$$r0_{entry\ j,d} = \frac{r_{entry\ j,d} - u_{entry\ j,d} \cdot r100}{1 - u_{entry\ j,d}}$$

Equation 65

Finally, the more preventive of the two criteria is selected as the definitive *r0* value:

$$r0_{j,d} = min\left(r0_{score\ j,d} - r0_{entry\ j,d}\right)$$

[83] Kee and Karwowski, "LUBA: An Assessment Technique for Postural Loading on the Upper Body Based on Joint Motion Discomfort and Maximum Holding Time."

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Equation 66

LUBA is thus used to modulate the angular sensitivity of the instantaneous calculation. The result is neither the LUBA index nor a direct reproduction of its scores, but a continuous adaptation based on proximity to the effective ROM.

15.2.3 Determining r0 for the lower limbs using isocomfort equations

LUBA does not provide tables for the hip, knee, and ankle. For the lower limbs, Ergoniza Intelligent therefore uses an alternative procedure based on the isocomfort equations[84] available for these body regions (see Section 11.2.1). These equations estimate perceived comfort *y* as a function of joint angle:

$$y_{j,d} = f_{j,d}(ang)$$

Equation 67

For each joint *j* and degree of freedom *d*, comfort in the neutral position is calculated first:

$$y0_{j,d} = f_{j,d}(0)$$

Equation 68

as is estimated comfort at the effective ROM limit:

$$yROM_{j,d} = f_{j,d}(ROM_{ef\ j,d})$$

Equation 69

The total decrease in comfort between the neutral position and the effective ROM is then calculated:

$$\Delta y_{j,d} = y0_{j,d} - ROM_{j,d}$$

Equation 70

The total decrease in comfort is used to generate a score equivalent to the maximum LUBA score. This equivalent score is normalized within the available set of degrees of freedom for the lower limbs:

$$S_{eq\ j,d} = 5 + 10 \cdot \frac{\Delta y_{j,d} - \Delta y_{min}}{\Delta y_{max} - \Delta y_{min}}$$

Equation 71

where Δy_{min} and Δy_{max} are, respectively, the smallest and largest decreases in comfort observed among the available lower-limb movements in the calculation context.

An equivalent angle for entry into the high-discomfort zone is also defined. This is done by calculating the angle at which a specified proportion of the total decrease in comfort is reached. The current implementation uses 75%:

$$f_{j,d}(\theta_{entry\ eq\ j,d}) = y0_{j,d} - 0.75 \cdot \Delta y_{j,d}$$

Equation 72

The value $Seq_{j,d}$ is used as the equivalent of $Smax_{j,d}$ and $\theta entry_{eq\ j,d}$ as the equivalent of $\theta entry_{j,d}$. From these values, *r0* is calculated using the same procedure described from Equation 61 through Equation 66.

This procedure incorporates the hip, knee, and ankle without directly extrapolating LUBA beyond its anatomical scope. The underlying logic remains consistent: the indicator continues to be based on proximity to the effective ROM, while the sensitivity of each movement is adapted according to the loss of comfort estimated by the isocomfort equations.

15.2.4 Operational age adjustment of effective ROM

The ROM used in this procedure is not necessarily the base ROM of the biomechanical model. When calculating instantaneous postures, Ergoniza Intelligent applies an operational age adjustment to the ROM used as a reference:

$$ROM_{ef\ j,d} = ROM_{base\ j,d} \cdot F_{edad\ j,d}$$

Equation 73

where:

- $ROM_{base,j,d}$ is the model's reference ROM for joint *j* and movement *d*;
- $Fedad_{j,d}$ is the age-adjustment factor;
- $ROMef_{j,d}$ is the effective ROM used solely to calculate relative proximity to the joint limit.

This adjustment neither changes the model's actual ROM nor prevents a particular posture from being adopted. It changes only the denominator used to calculate *rj,d*. Thus, for an older person, the same angle may represent greater relative proximity to the functional limit.

[84] Kee and Karwowski, "Joint Angles of Isocomfort for Female Subjects Based on the Psychophysical Scaling of Static Standing Postures"; Kee and Karwowski, "The Boundaries for Joint Angles of Isocomfort for Sitting and Standing Males Based on Perceived Comfort of Static Joint Postures."

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The age factors used for the shoulder, elbow, hip, knee, and ankle were derived from the ROM reference values published by Soucie et al.[85], using ages 20–40 as the reference group. Factors greater than 1 are applied to the under-20 group when the data show a higher mean ROM, and factors below 1 are applied to the over-40 group when the data show a reduction in ROM.

For the neck and trunk, for which the data from Soucie et al. do not provide equivalent coverage, operational factors are based on specific studies of age-related loss of cervical and lumbar mobility. Kuhlman et al.[86] describe different reductions in cervical ROM for flexion, extension, lateral flexion, and rotation in older populations, while Intolo et al.[87] show that the effect of age on lumbar ROM is not uniform across movement planes.

The literature also indicates that age-related loss of mobility is joint-specific. Flexitest results[88] from a large sample show that the shoulder and trunk tend to lose proportionally more mobility than the elbow and knee, supporting the use of different age adjustments for different body regions.

Based on these studies, Ergoniza Intelligent uses the following factors:

	Age range		
Region	<20	20-40	>40
Shoulder	1.01	1.00	0.97
Wrist	1.02	1.00	0.95
Elbow	1.01	1.00	0.98
Hip	1.06	1.00	0.91
Knee	1.02	1.00	0.97
Ankle	1.11	1.00	0.90
Neck - flex.	1.02	1.00	0.94
Neck - ext.	1.02	1.00	0.84
Neck - lat.	1.02	1.00	0.89
Neck - rot.	1.02	1.00	0.88
Trunk - flex.	1.03	1.00	0.90
Trunk - ext.	1.03	1.00	0.88
Trunk - lat.	1.03	1.00	0.90
Trunk - rot.	1.00	1.00	0.98

Table 7: Operational factors for age adjustment of effective ROM.

These values should not be interpreted as normative or universal, but as an ergonomic parameterization based on the literature. They may be adjusted in future versions as new experimental data become available.

[85] J. M. Soucie et al., "Range of Motion Measurements: Reference Values and a Database for Comparison Studies," *Haemophilia* 17, no. 3 (2011): 500–507, doi:10.1111/j.1365-2516.2010.02399.x.

[86] Kurt A. Kuhlman, "Cervical Range of Motion in the Elderly," *Archives of Physical Medicine and Rehabilitation* 74, no. 10 (1993): 1071–79, doi:10.1016/0003-9993(93)90064-H.

[87] Pattariya Intolo et al., "The Effect of Age on Lumbar Range of Motion: A Systematic Review," *Manual Therapy* 14, no. 6 (2009): 596–604, doi:10.1016/j.math.2009.08.006.

[88] Hugo Baptista De Oliveira Medeiros, Denise Sardinha Mendes Soares De Araújo, and Claudio Gil Soares De Araújo, "Age-Related Mobility Loss Is Joint-Specific: An Analysis from 6,000 Flexitest Results," *Age* 35, no. 6 (2013): 2399–2407, doi:10.1007/s11357-013-9525-z.

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