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Determining the Acceptable Limits of Head Mounted Loads

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ABSTRACT

This report reviews ergonomic and biodynamic factors related to the introduction of head mounted loads for military land environments. It presents an overview of current knowledge pertaining to the assessment of head mounted load limits. Limitations in the predictive ability of existing modelling and measurement techniques are discussed. The report introduces a novel biodynamic model, currently under development by DSTO, designed to overcome these limitations. LOD's Human Biodynamics Engine is a realistic model of the human body motion within a dynamic environment. The model is unprecedented in its representation of the full, distributed, three-dimensional neck and spine dynamics and in its incorporation of reflexive and cerebellar-like control of spinal motion. The model is used to investigate specific considerations of head mounted load design and their use in an operational context.

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Determining the Acceptable Limits of Head Mounted Loads

Executive Summary

This report reviews ergonomic and biodynamic factors related to the introduction of head mounted loads (HML) to military land environments. Through its review of current literature, the report summarises the effect of the mass distribution of HML on physiological response and endurance, which are the basis for existing estimates of acceptable helmet and HML design limits. The ability of each of the reported studies to reach definitive design standards rests on the current limitations in the measurement and modelling of musculoskeletal torques and forces in live subjects under realistic dynamic situations.

The addition of mass and its distribution in relation to the pivot points of the spine are the basic attributes by which HMLs can be distinguished. Tied up with these attributes are therefore the factors that increase the risk of injury to the wearer. The moment of inertia is a mechanical quantity that reflects the mass distribution of a HML at each axis of rotation. The basic ergonomic recommendation is to choose the HML that produces the smallest overall moment of inertia with respect to the neck joints. In simple practical terms this means — choose the HML which has the smallest mass, is the most symmetrically balanced and aligned to the natural head's centre of mass, and is closest to the head (e.g. with the smallest diameter in the case of a helmet).

LOD's Human Biodynamics Engine (HBE) is a realistic model of human body motion within a dynamic environment. The HBE attempts to model whole skeleton forward and inverse dynamics with 300+ degrees of freedom driven by the same number of equivalent muscular actuators (each with its own excitation-contraction dynamics). It includes two levels of neural-like control (stretch-reflex and cerebellar stabilizer). The spine has been modelled as a chain of 25 loosely coupled rigid bodies (technically, a chain of 25 SE(3) groups), using either Hamiltonian or Lagrangian formalisms. Rotational dynamics is represented as an "active" neuro-muscular stretch-reflex with cerebellum-like control. Translational dynamics is represented as a "passive system" including discs, tendons and ligaments as a nonlinear spring-damper system. The model is unprecedented in its representation of the full, distributed, three-dimensional neck and spine dynamics and in its incorporation of reflexive and cerebellar-like control of spinal motion. The model is used to investigate specific considerations of head mounted load design and their use in an operational context.

The HBE's detailed modeling of the spine's rotational dynamics raised questions regarding traditional approaches to the estimating of injury risk. Cervical spine injuries are normally classified based on the principal loading applied to the cervical spine, including tension, compression, bending and shear. While these factors do cause severe, fracture-type, spinal injuries, they do not explain the vast majority of neck and back injuries. More detailed biomechanical analysis, performed using the HBE, reveals the deeper mechanism for joint injuries in general, and for cervical-spine injuries in particular. According to this analysis, the main risk factor for the spinal injuries is high

torque-jolt (TJ) in the spinal joints, which has two components: rapid, or "jerky" angular accelerations/decelerations (e.g. a rapid change of direction during a sport, rapid deceleration during a vehicle crash), and significant moments of inertia.

TJs can be considered at the *global* (whole body) level and the *local* (intervertebral) level. At the local level the greatest risk of injury is when the TJ in a single joint is much higher than the TJs in its neighbouring joints. This distinction between local and global TJ helps to explain the body's tolerance for large global forces. Pilots are able endure high G-force without sustaining an injury when this force changes smoothly, i.e., without high local TJs. Similarly, we are able to tolerate relatively high whole body jolts without injury when these forces are sufficiently distributed at a local level to avoid high TJ gradients at the vertebra.

All other biomechanical conditions are considered as secondary risk factors. They include (i) proximity to anatomical joint limits, i.e., passive strain on joint ligaments, tendons and muscles at the edge of natural anatomical motion; (ii) muscular insufficiency and fatigue, and (iii) the properties of HML, including mass and its eccentricity from the head's centre of mass.

If we accept high TJs as being the underlying mechanism of injury, the degree to which the loaded head and neck is in dynamic motion influences the degree to which the properties of the HML should be prioritised. A HML worn in a dynamic situation, e.g. by a soldier walking or running, adds to the forces of rotation in whichever direction the cervical spine joints of the neck are rotating. A counterbalance for a HML, despite reducing the inertial moment tending towards forward rotation, imposes an additional mass thus increasing the overall inertial movement in each of the five other directions of rotation as well as compressing the neck, thus raising the risk of high TJs and injury. Given the human adaptation of strong neck and shoulder muscles to support the lifting and lowering of the head, we are better suited to tolerating smaller forward mounted loads on the head than to the addition of loads that increase forces acting on head rotation and lateral movement. For dynamic situations, muscle strength and endurance is a key contributor to reducing the risk of injury.

In predominantly static situations, e.g. seated whilst observing fixed position where balance of the load is achievable for prolonged periods, the use of a counterbalance to reduce the eccentricity of the HML may be more clearly justified against the contribution of the additional mass. In this case, muscle strength and endurance is less of a factor for reducing risk, as the balanced static system requires virtually no muscular energy.

Regarding the use of HML whilst seated in moving vehicles or during fire and movement, the risk of cervical spine injuries can be associated with three main components: (i) high TJs of both vehicle and body motion; (ii) slippage of the HML; and (iii) the mass distribution of HML (inertia moments per axis of neck rotation).

If the HML is predefined, the recommendation is to reduce HML slippage and to minimize vehicle and body TJs. In vehicles, a reduction of the currently recommended speed of 60 km/h for firm roads and 50 km/h for cross-country to 40 km/h and 30 km/h respectively would represent significant reduction of risk arising from cervical spine injury.

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Contents

Glossary

3D	three-dimensional
AP	action potential
CG	centre of gravity
CoM	centre of mass
DOF	degrees of freedom
FE	finite element
F-t	force-time
F-v	force velocity
G	gravity
HBE	Human Biodynamics Engine
HML	head mounted load
HSD	head supported display
LOD	Land Operations Division
MI	moment of intertia
MVC	maximal voluntary contraction
N	Newton
NVG	night vision goggles
TJ	torque-jolt
USAARL	US Army Aeromedical Research Laboratory
WBV	whole-body vibration

1. Introduction

Emerging concepts of the future infantry soldier depict a skilled team member utilising the advantages of technology within an information rich battlefield to apply force with increasing efficiency. The introduction of the associated equipment places additional loads on the user, both physical and mental. Given the inflexible boundaries of human physical capability, the functional benefits offered by this equipment are achieved at a cost to the physical capability of the wearer. As the infantry soldier must carry the majority of their equipment onto the battlefield, the physical load associated with a new technology must be balanced against operational advantage.

Relatively recent technological advances have seen the introduction of night vision goggles (NVG) and head mounted displays within an infantry context. These important advances in capability come at the cost of raising the head mounted load (HML) on the individual. The field of biomechanics has struggled to establish clear standards for human loading due to a range of barriers to investigation including subject welfare, system complexity and measurement limitations. This report draws together current knowledge of the issues associated with the introduction of HMLs for the infantry soldier and describes ongoing research to accurately predict the biomechanical consequences of human loading under a range of operational environments.

2. Background

2.1 Principles of Head Mounted Loading

Research related to the impact of HMLs is focused, predominantly, on the cervical spine (the neck). The cervical spine fulfils three central functions [1]:

1. Supporting the weight of the head;
2. Allowing the head to move in a wide range of directions,
3. Protecting the nervous system, i.e., the spinal cord.

The construction of the cervical spine allows each of these functions to be fulfilled. Any deficiency in the cervical spine structures may be reflected in an inability to satisfactorily fulfil these functions. The cervical spine is a mechanical structure [2] constructed from a range of different biological tissues, organised in different structural units. These include the cervical muscles, spinal ligaments, seven vertebrae, intervertebral discs, and zygapophyseal joints. The extent to which each structure is responsible for adequate cervical spine function varies in different situations as the intrinsic load carrying capacities of these components are altered by different postures and external loads. Of prime concern is the capacity of the system structure to provide mechanical stability in the range of situations to which it may be asked to respond [3]. Chronic or acute deformation or damage of these structures may undermine this mechanical stability.

The tolerance of the cervical spine to the strain of physiological loads will depend on the initial and end orientations of the head, cervical spine and torso, the material properties of the spinal tissue, the geometry of the vertebrae and their buckling behaviour, the direction and magnitude of the force applied, the inertial forces and the level of muscle activity present [4]. Research into the practical limits of HMLs has mainly been performed in aeromedical research laboratories. Aviators flying rotary-wing aircraft are exposed to whole-body vibration (WBV), transmitted primarily through the seating system causing musculo-skeletal stress to the back and neck [5,6]. These stresses are aggravated when the head is further loaded with HMLs.

The effects of 15 helmet configurations on the fatigue of neck muscles were examined by Phillips and Petrofsky [7]. The authors utilized a helmet simulator by adding three weights at five different centre of gravity (CG) locations on the helmet. Six male subjects performed 30 minutes of right and left lateral rotation of the head while wearing one of the helmet configurations. Each subject then pulled against a load equivalent to 70 percent of his maximal voluntary contraction (MVC) and sustained this exertion until fatigued. They found, in general, that endurance time was sensitive to weight and CG location of helmet mass. They recommended that, for a three-pound helmet, the optimal CG location should be 5 cm in front of head-neck CG and for a nine-pound helmet, the helmet CG should be behind head-neck CG.

A series of studies has been conducted at the US. Army Aeromedical Research Laboratory (USAARL) to evaluate the effects of HML mass properties on biomechanical, physiological and performance responses of male pilots (see [8,9,10]). The authors found that the weight-moment of HMLs worn by male aviators should not exceed 82.8 ± 22.9 Ncm. Butler [8] showed a significant increase in head pitch acceleration response when the total head-supported load exceeded 83 Ncm relative to the atlanto-occipital complex. Lantz [9] showed significant changes in electromyography responses to head supported display (HSD) loading under WBV. Alem et al. [10] studied performance of male pilots under long exposure (up to 4 hours) to WBV and under four HSD configurations. They demonstrated that the subject's reaction time to a randomly appearing target increased as the weight moment of the helmet increased beyond 78 Ncm.

Butler and Alem [11] exposed twelve U.S. Army volunteer aviators to 4 hours of WBV, similar to that found in a UH-60 helicopter, while wearing four different helmets. Helmet torques, as calculated at the point where the head connects to the spine, ranged from a standard aviator helmet to a helmet with a chemical mask and a night vision goggle. Head motion was measured using a 3D active infrared marker system attached to a fixture held in the subject's teeth. Results showed no significant differences ($p<0.05$) in head pitch motion over time for helmets, but significant differences among helmet torques. These results support the existing recommended helmet design of limiting the added helmet torque to 90 Ncm for long-duration helicopter flights.

Barazanji and Alem [12] defined a safe range of weights and centres of mass of HSDs that could be tolerated by female helicopter pilots without affecting their health or degrading their performance. They exposed twelve subjects to whole-body vibration while wearing an HSD with various mass properties, recording biomechanical head acceleration, neck muscle activities, and performance responses. Head pitch, anterior-posterior, and axial accelerations were measured for 12 different helmet configurations during sinusoidal vertical vibration having a magnitude of 0.45 m/s^2 and frequencies swept from 2 Hz to 17 Hz at the rate of 0.25 Hz/sec. Obtained results indicated that head pitch and axial acceleration levels for female subjects were lower than those for their male counterparts. They found that negative loading (spine tension) had a detrimental effect on females but not on males. Their results also showed differences in magnitude of head pitch acceleration between weight moments higher and lower than 91.3 ± 28.6 Ncm, compatible to that recommended for their male counterparts (82.8 ± 22.9 Ncm). Based on the biomechanical response alone, they recommended that the design criteria of HML mass properties should not be gender sensitive.

Barazanji et al. [13] have proposed the operational guidelines for male aviators (see Figure 1), with neck torques defined as "biomechanical response" of 83 Ncm and "vigilance response" of 78 Ncm (not included the Figure 1).

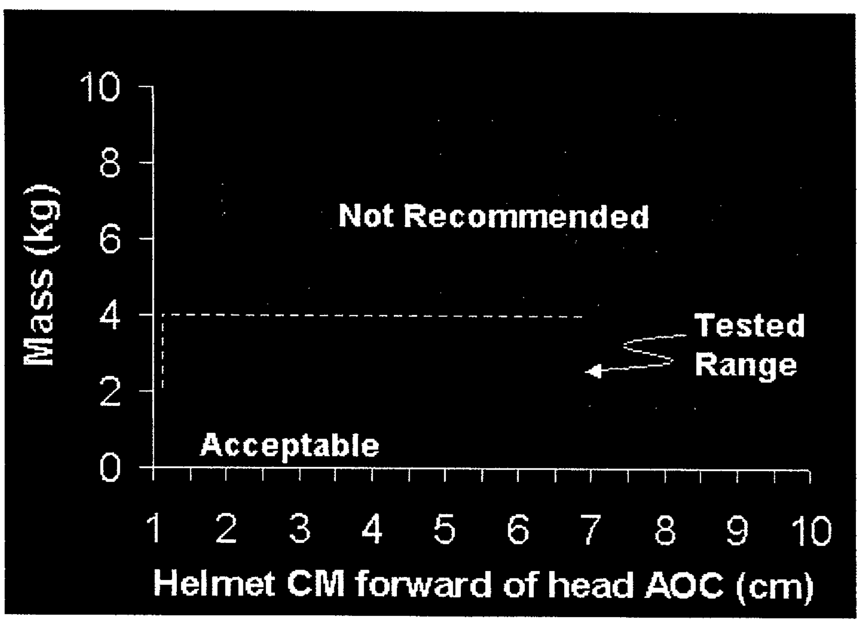

Figure 1. Operational guidelines for male aviators, proposed by Barazanji et al. [13]

2.2 Biomechanical Modelling

Mathematical modelling has long been recognised as a valuable tool to simulate and analyse the mechanical behaviour of biological structures (see [14]). Typically, three kinds of mathematical models of the human cervical spine have been reported in the literature to describe its dynamic behaviour in impact situations (see [15]): two pivot models, discrete parameter models and Finite Element (FE) models.

FE models, although currently fashionable within the biomechanics and biomedical engineering communities (supported by powerful and expensive software packages), are essentially *translational,* and predominantly either *static* or *linear.* Because of these characteristics, they do not have a real place in the *many degrees of freedom, highly nonlinear, rotational-joint* human *dynamics,* as considered in this report (in the same way as they are not applicable in humanoid robotics).

Pivot models describe the head/neck as a series of mechanical segments linked at pivot points. Two pivots are the simplest and represent the head and torso as rigid bodies connected by a rigid or extensible neck-link. The behaviour of the neck is combined as head-neck and neck torso pivots. However, the cervical spine motion is so complex that it cannot be replicated in two segment models connected by a pivot (see [16]).

Discrete parameter models are claimed to have more anatomical validity than pivot models [14]. They include head and vertebrae as rigid bodies connected by massless spring-damper mechanisms representing intervertebral soft tissues. Discrete parameter models have been reported as an efficient and effective method for the study of head/neck kinematics (see [17]). This approach, whilst not able to predict strain fields and failure modes of individual spinal components, describes the kinematics of the neck well. Tien and Huston [16] discussed issues associated with the development of the three and nine point parameter models to describe the head and neck biomechanics following whiplash trauma. Again, these elementary translational models are still too simplistic to reflect rotational-joint dynamics, while the spring-damper mechanisms cannot simulate muscular excitation and contraction dynamics.

The present report addresses ergonomic and biodynamical factors related to choice and acceptable limits of HML. The analysis of these limits has been performed from the perspective of the Human Biodynamics Engine (HBE), a realistic biomechanical model of the human body dynamics, currently under development in LOD (see [18-23,32-33]).

The HBE is the fourth kind of a human body model. The HBE uses non-linear kinematics-dynamics-control approach of modern humanoid robotics. It includes several hundred controlled degrees of freedom modelling all existing rotations and translations, realistic distribution of all the body masses, muscular excitation-contraction dynamics, and hierarchical cerebellum-like control. The complexity of the HBE model is unprecedented in the biomechanics literature, as well as its proximity to the real human anatomy and physiology.

3. Ergonomic Analysis of Head Mounted Loads

3.1 Moment of Inertia

The addition of mass and its distribution in relation to the pivot points of the spine are the basic attributes by which HMLs can be distinguished. Tied up with these attributes are therefore the factors that increase the risk of injury to the wearer. The moment of inertia (MI) is a mechanical quantity (one entry in an *inertia matrix*) that reflects the mass distribution of a HML per axis of rotation. This means all objects have three MIs, with respect to say the X, Y, and Z axes of rotation.

3.1.1 Why is moment of inertia important?

MI is the key parameter in the ergonomic choice of HML because of the rotational form of the crucial Newton's law, which says: **the total torque acting around a single joint axis equals MI multiplied by angular acceleration.** The angular acceleration is produced by human movement, and the result of it, which actually injures the neck, is the total torque. This includes muscular, inertial and gravity torques as well as other internal torques, such as tendon elasticity and joint damping, and also external torques, like various kinds of crashes. Neck muscles contract to produce the compensating torque for inertial, gravity and other torques. Adequate compensation occurs only when the muscle response is fast enough and strong enough to match the onset of external torques. Speed and strength are both diminished by fatigue, raising the risk of injury over prolonged periods of loading.

Therefore, the basic ergonomic recommendation is to choose the HML, which produces the smallest overall moment of inertia with respect to the neck joints. In simple practical terms this means - choose the HML which:

(a) Has the smallest mass;
(b) Is the most symmetrically balanced and aligned to the natural head's centre of mass; and
(c) Is closest to the head (e.g. with the smallest diameter in the case of a helmet).

3.1.2 Determining the moment of inertia.

Usually the oversimplified model is used for the MI: the whole mass of an object is considered to be concentrated in a single point, the so-called "centre of mass" (CoM, not "centre of gravity", as is commonly used, given that we are dealing primarily with inertial forces rather than gravity), and then the MI is calculated as one-half of its **mass × distance^2** from the rotation pivot point (in our case an imaginary point located somewhere among the neck vertebrae, around which the head rotation is supposed to happen). The current state of biodynamics forces the use of this oversimplified approach. To make it user-friendly, we use the artificial parameter "eccentricity", which distinguishes the natural head's centre of mass from the loaded head's centre of mass.

In reality, the mass of a HML is not concentrated in a single point, but rather distributed (not to mention the sliding among the HML-components). Mathematically speaking, instead of a single, average point, we have an integral continuum, which is a much more complicated object; the mass distribution of any object is represented by the so-called inertia tensor with nine components, each calculated as a volume integral dependent on the mass density and various squared distances. Reducing this complex geometrical object to the simple **mass × distance**^{2} is obviously oversimplified, with or without eccentricity; it is something like modelling the motion of a car by a single moving particle.

Particularly in case of a spherical helmet, we might say that the eccentricity of the head-helmet CoM from the pure head's CoM is the key parameter, but the radius of this sphere also counts – the bigger the radius, the bigger the MI with respect to the axis aligned with the radius. In the case of NVGs the inapplicability of the concentrated model is obvious.

3.2 Cervical Spine Injuries: Injury Mechanism and Risk Factors

3.2.1 Mechanism of Cervical Spine Injuries

Cervical spine injuries are normally classified based on the principal loading applied to the cervical spine, including tension (longitudinal extension), compression (longitudinal contraction), bending and shear [24] [25].

- *Vertical compression* (Jefferson fracture, multipart atlas fracture, vertebral body compression fracture, burst fracture).
- *Compression-flexion* (vertebral body wedge compression fracture, hyper-flexion sprain, unilateral facet dislocation, bilateral facet dislocation, teardrop fracture).
- *Compression-extension* (posterior element fracture).
- *Tension* (occipito-atlantal dislocation).
- *Tension-extension* (whiplash, anterior longitudinal ligament tears, disk rupture, horizontal vertebral body fracture, hangman's fracture, teardrop fracture).
- *Tension-flexion* (bilateral facet dislocation).
- *Torsion* (rotary atlanto-axial dislocation).
- *Horizontal shear* (anterior and posterior atlantoaxial subluxation, odontoid fracture, transverse ligament rupture).
- *Lateral bending* (nerve root avulsion, transverse process fracture).
- *Other fractures*.

Now, while these factors do cause severe, fracture-type, spinal injuries, they do not explain the vast majority of neck and back injuries. In the next subsection we propose a new look at the mechanism of the majority of spinal injuries.

3.2.2 Risk Factors for Cervical Spine Injuries

More detailed biomechanical analysis, performed using HBE, reveals the deeper mechanism for joint injuries in general, and for cervical-spine injuries in particular. According to this analysis, the main risk factor for the spinal injuries is **high torque-jolt (TJ)** in the spinal joints, which has two components:

1. Rapid, or "jerky," angular accelerations/decelerations (eg. a rapid change of direction during a sport, rapid deceleration during a vehicle crash), and
2. Significant moments of inertia, i.e.,

   Torque-jolt = Angular Acceleration Jerk × Inertia Moment

   (See Appendix A for technical details).

Regarding the local spinal TJs, the worst-case scenario is to have high TJ in a single intervertebral joint with much lower TJs in the neighbouring joints. This would, almost certainly, cause a spinal injury in that particular spot. Also, we can endure relatively high G-force or G-acceleration without sustaining an injury – if this force/acceleration changes smoothly. Finally, we can even have high external translational jolt (not to mention high force/acceleration) and yet have a reasonably manageable TJs in the joints (and relatively low risk of injuries) in the case of small joint lever arms. In the case of the spine, each particular intervertebral joint has a small lever arm. However, the full-distributed spine (including 25 movable intervertebral joints) has a big lever arm, which produces a big overall TJ.

Therefore, to reduce the overall risk of cervical spine injuries, we need to prevent both high local TJs and high overall TJs.

All other biomechanical conditions are considered as **secondary risk factors**. They include:

1. **Proximity to anatomical joint limits,** which means passive strain on joint ligaments, tendons and muscles at the edge of natural anatomical ranges (eg. when looking over your shoulder or bending to pick an object off the floor).
2. **Muscular insufficiency and fatigue**, and
3. **Properties of HML**, including
   - (i) Mass [kg],
   - (ii) Eccentricity [cm], which is a distance between HML CoM vs. anatomical head CoM, and
   - (iii) Diameter [cm].

Here we consider high TJs to be the underlying mechanism of injury. The level of TJ that can be tolerated without injury depends on the nature of the joint and its proximity to its limits of rotation. A weaker joint near the edge of its full range of rotation will be injured by a lower level of TJ than would a stronger joint in a relaxed position. The mitigation of this injury risk factor is to seek to ensure that external forces and required body movements are maintained within known bounds of joint rotations and that these bounds are reduced to accommodate any additional loading placed on the individual.

The muscle groups controlling the rotation of our various joints are critical in resisting and dissipating those forces that would push each joint beyond its limits of rotation. Here we propose that it is the instantaneous application of force rather than the more gradual massive loading of the joint that must be resisted and dissipated by the joint and its associated muscles to avoid injury. This can be clearly seen in cases of injury where joints are forced in a direction in which the muscle groups are unsuited to, and incapable of, countering imposed forces e.g. acute lateral flexion of the knee. Given this vital role of the muscle groups associated with each joint, muscle strength and the effects of fatigue are important factors for the mitigation of TJ related injury. The mitigation for muscular insufficiency and fatigue is one of training to build muscle strength, rest cycles and the work regime to avoid insufficient resistance and dissipation of force at the extremes of joint motions.

The final factor influencing the risk of injury is the design of the HML itself. If we accept high TJs as being the underlying mechanism of injury, the degree to which the loaded head and neck is in dynamic motion influences the degree to which the properties of the head mounted load should be prioritised. A HML worn in a dynamic situation, e.g. by a soldier walking or running, adds to the forces of rotation in whichever direction the cervical spine joints within the neck are rotating. A counterbalance for an NVG, despite reducing the inertial moment tending towards forward rotation, imposes an additional mass, thus increasing the overall inertial movement in each of the five other directions of rotation as well as compressing the neck thus raising the risk of high TJs and injury. Given the human adaptation of strong neck and shoulder muscles to support the lifting and lowering of the head, we are better suited to tolerating smaller forward mounted loads on the head than to the addition of loads that increase forces acting on head rotation and lateral movement. For dynamic situations, muscle strength and endurance is a key contributor to reducing the risk of injury.

In predominantly static situations, e.g. seated whilst observing fixed position where balance of the load is achievable for prolonged periods, a counterbalance to reduce the eccentricity of the HML may be more clearly justified against the contribution of the additional mass. In this case, muscle strength and endurance is less of a factor for reducing risk, although it will continue to play a part for prolonged operations, as the balanced static system requires virtually no muscular energy to sustain.

Specifically, regarding *seating in the moving vehicles* as well as *moving and marksmanship using various firing positions*, the risk of cervical spine injuries contains three main components:

1. High TJs of both vehicle and body motion (all sorts of impacts in the vehicles; impact landings and jumps while moving and firing).
2. Slippage of the HML.
3. Mass distribution of HML (inertia moments per axis of neck rotation).

If the HML is predefined, the recommendation is:

1. To ensure that the HML does not slip on the head;
2. To avoid, or at least minimize vehicle and body TJs:
   (i) In vehicles, do not exceed the speed of 40 km/h on the firm road (see Figures 2-7 for the effect, measured by local TJs, of a crash into a wall with different driving speeds and head mounted loads); and
   (ii) In vehicles, do not exceed the speed of 30 km/h across rough terrain (see Figures 8-9 for the effect, measured by local TJs, of driving with different speeds and head mounted loads across rough terrain).

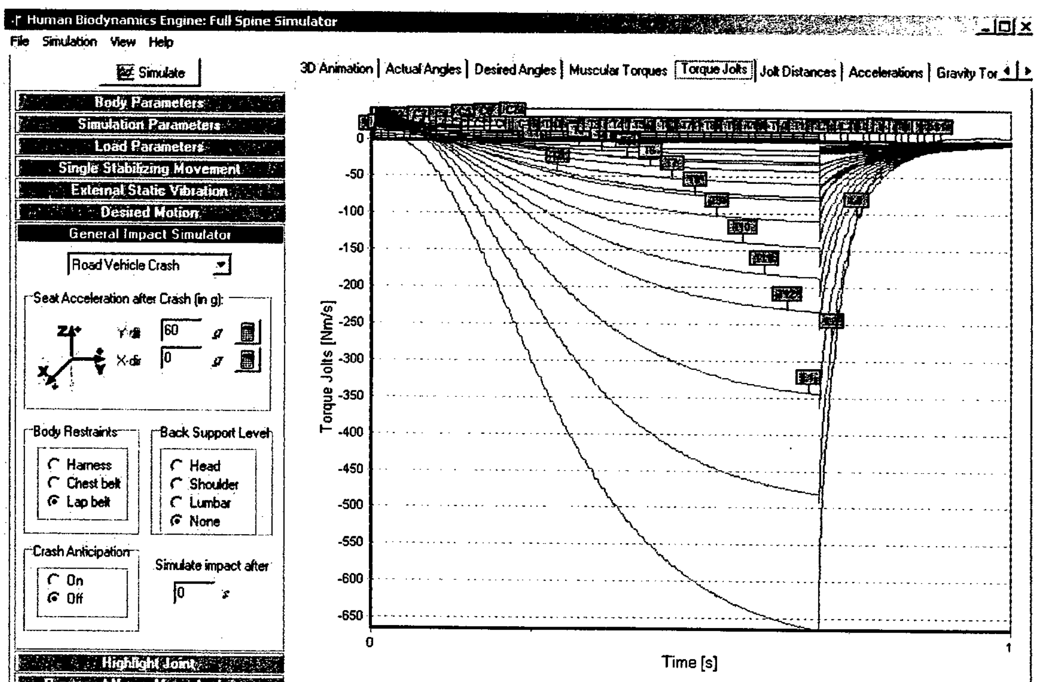

Figure 2. Torque-Jolts calculated for a crash into a wall from driving speed of 60 km/h in a sitting position without a chest belt and without a HML

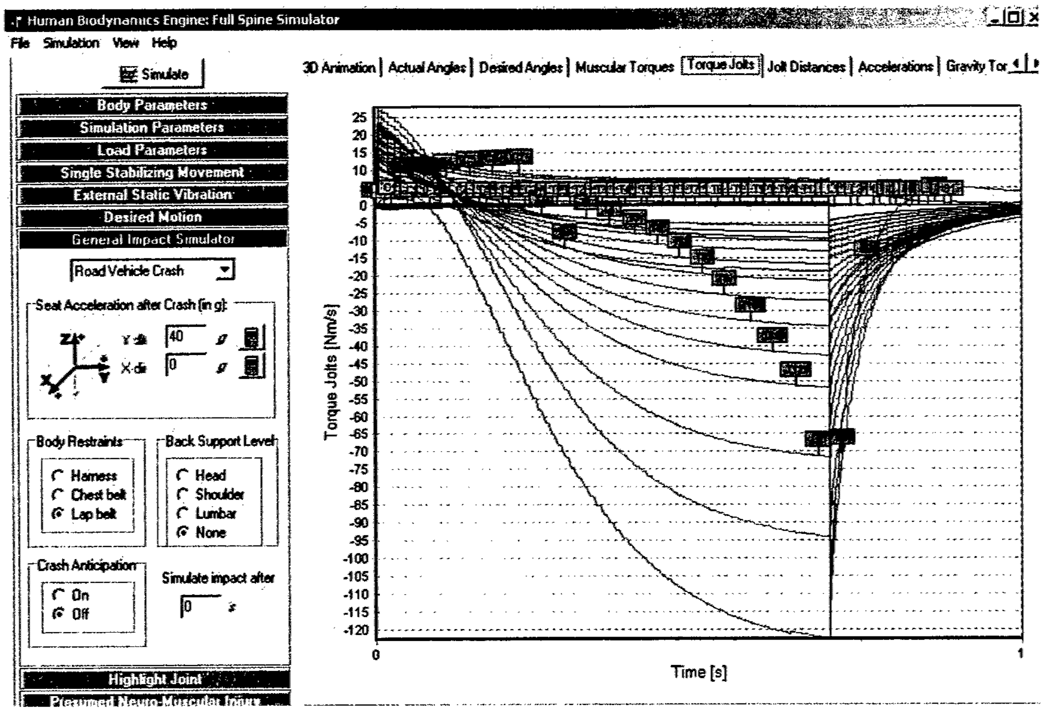

Figure 3. Torque-Jolts calculated for a crash into a wall from driving speed of 40 km/h in a sitting position without a chest belt and without a HML

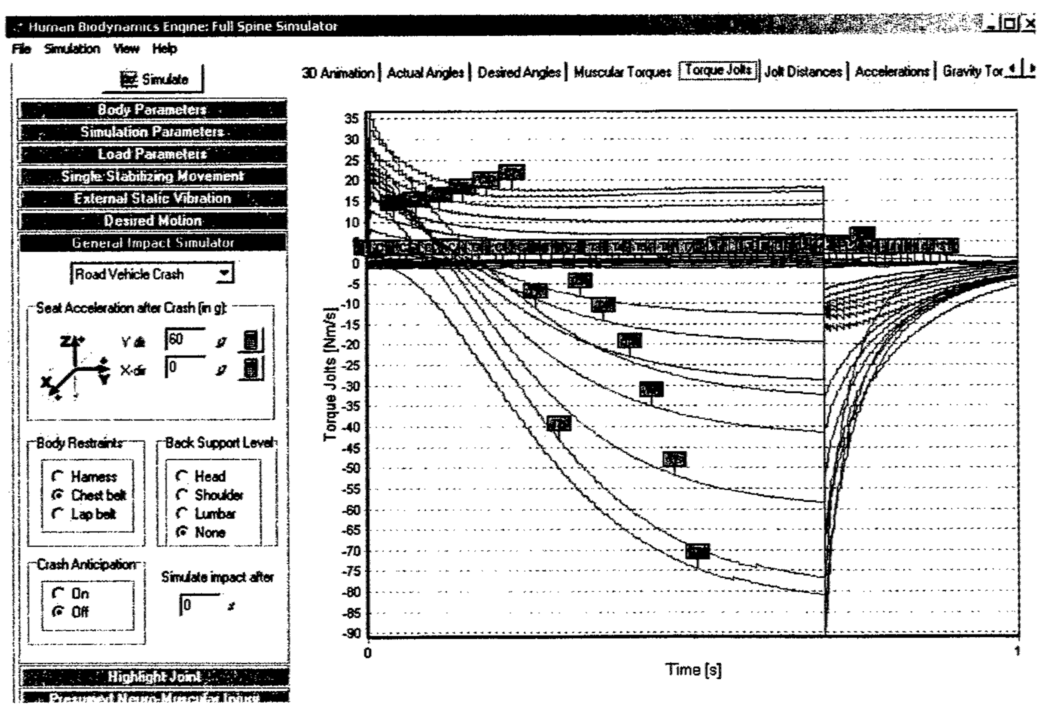

Figure 4. Torque-Jolts calculated for a crash into a wall from a driving speed of 60 km/h in a sitting position with a chest belt and without a HML

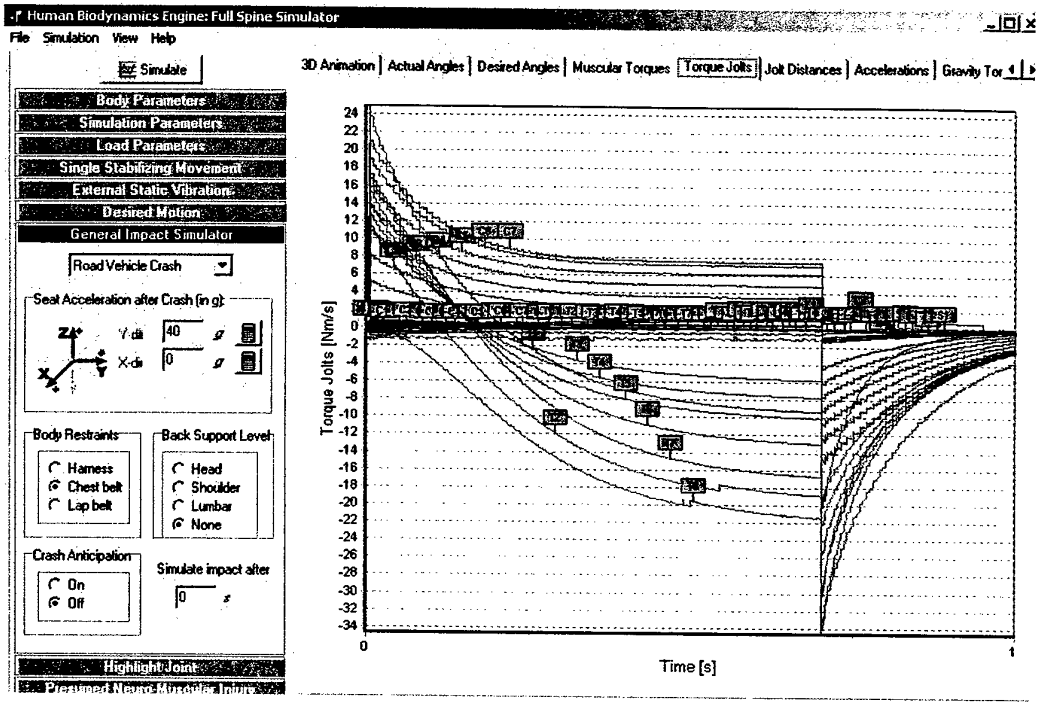

Figure 5. Torque-Jolts calculated for a crash into a wall from a driving speed of 40 km/h in a sitting position with a chest belt and without a HML

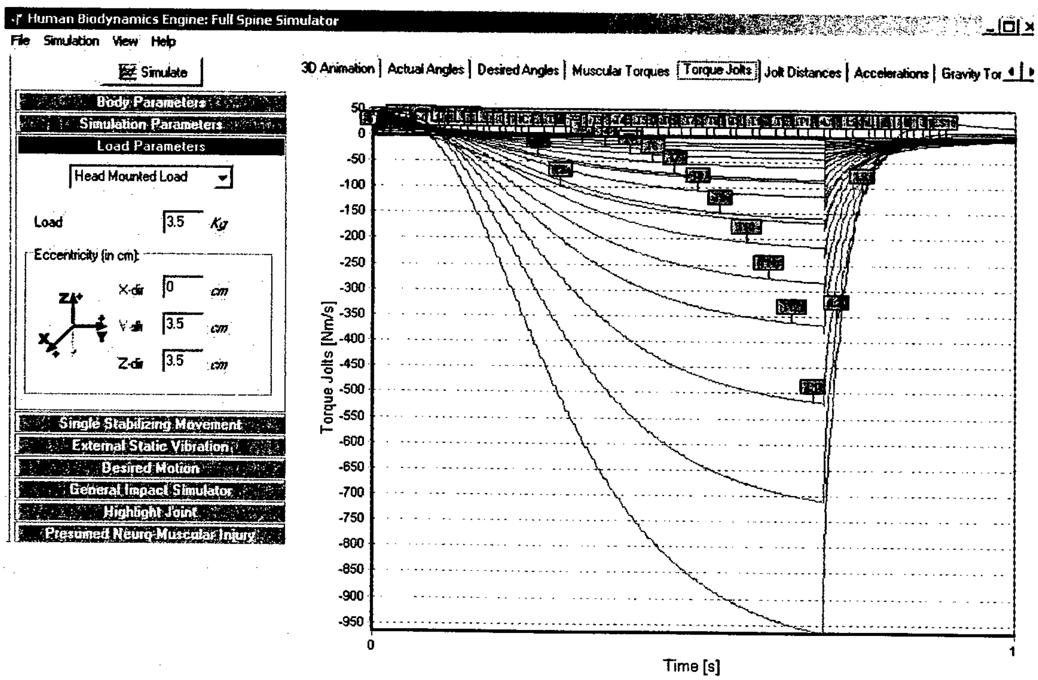

Figure 6. Torque-Jolts calculated for a crash into the wall from a driving speed of 60 km/h in a sitting position without a chest belt and with a 3.5 kg HML

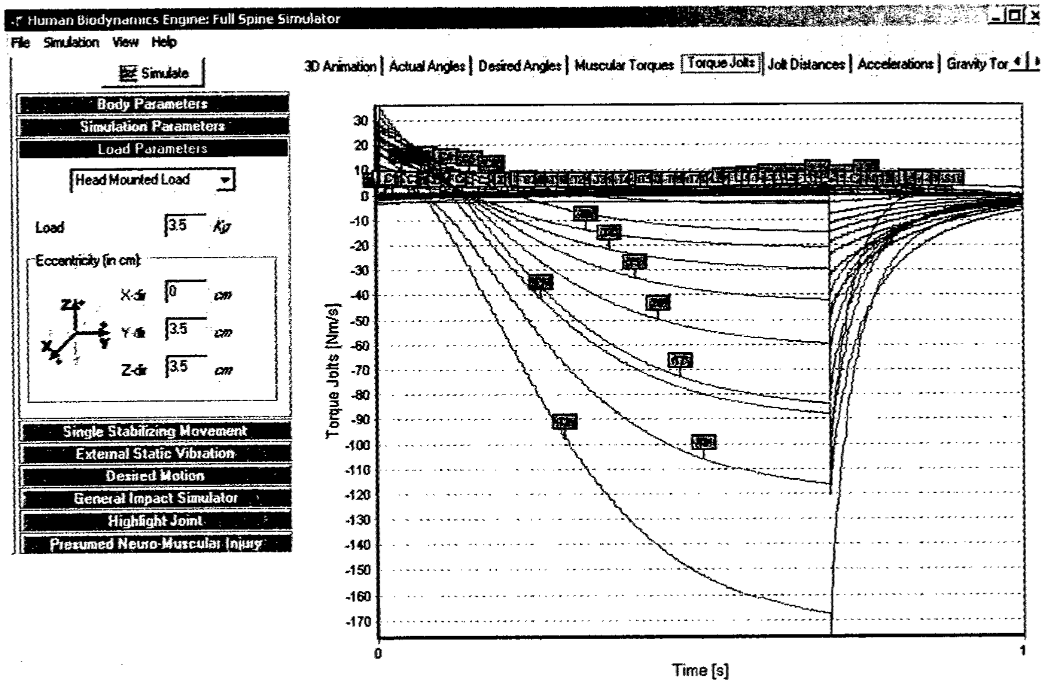

Figure 7. Torque-Jolts calculated for a crash into the wall from a driving speed of 60 km/h in a sitting position with a chest belt and with a 3.5 kg HML

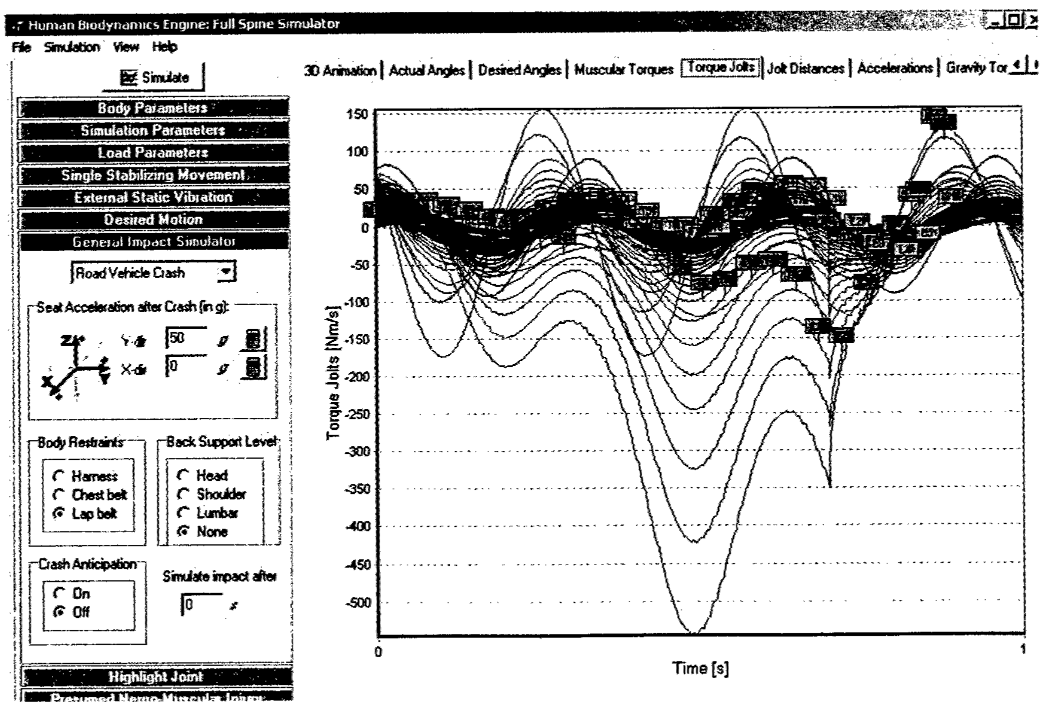

Figure 8. Torque-Jolts calculated for a driving speed of 50 km/h across rough terrain in a sitting position without a chest belt and with a 3.5 kg HML

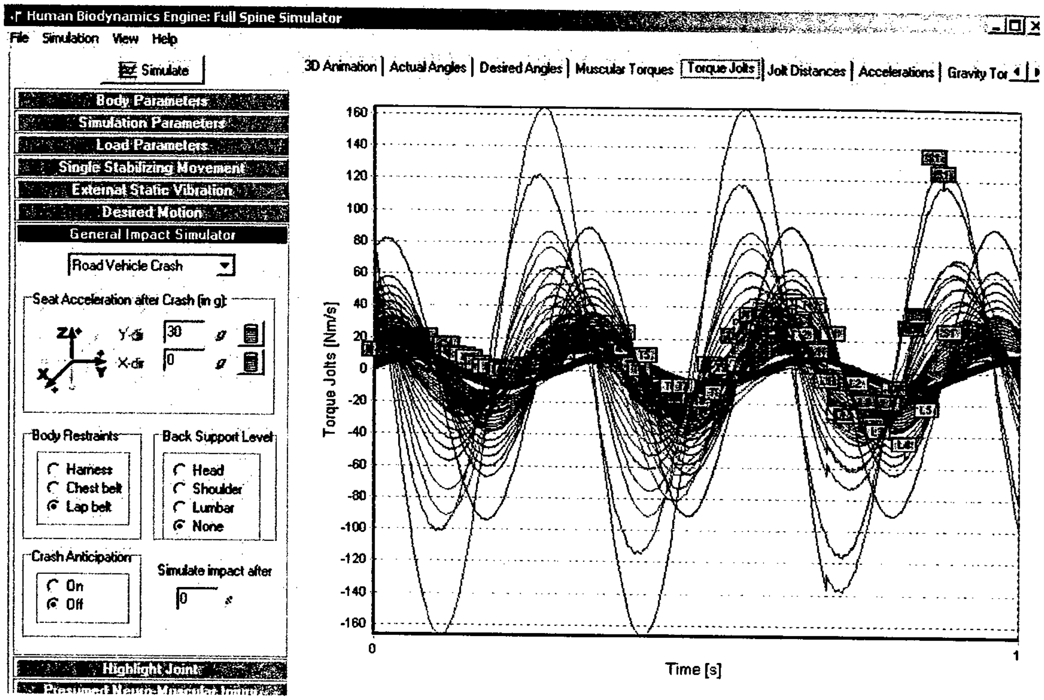

Figure 9. Torque-Jolts calculated for a driving speed of 30 km/h across rough terrain in a sitting position with a chest belt and with a 3.5 kg HML

4. Mathematical Modelling

4.1 Mathematical Modelling Versus Field Testing

Although the principles of biomechanics and HMLs offer guidance for reducing wearer risk when selecting designs for systems that impose a physical load on the user, little is known about the actual limits of operation and endurance that can be safely maintained in an operational environment. Answering such questions presents a significant challenge to the field of biomechanics. The approach advocated here is one of modelling to determine the interactions of loads and the dynamic environment of the human spine.

A simulator based on a valid mathematical model has the value of many field tests. It has the further advantage of being much cheaper and faster than a single realistic field test. Regarding mathematical models, in practice you find several cases:

1. A pure mathematical model, represented by dynamical equations, has the value of a physical law, which means it is absolutely true in its domain of validity

(classical equations for classical physics, quantum equations for quantum physics, etc.).

2. A mathematical model with validated empirical parameters[1] has a value which approaches the deductive model of point (1), provided the parameters are measured using a physically-valid measurement procedure (with a measurement error less than 5%). This is extremely rarely achieved in physiological measurements, and never in psychological measurements.

3. A mathematical model with non-validated empirical parameters (i.e., parameters that cannot be effectively measured) has a value that depends on the level of its validation (e.g. a biodynamic simulator can have *kinematic validation*, i.e. it predicts correct angular velocities and accelerations, without having *dynamic validation*, i.e. non-validated prediction of forces and torques simply because there are no relevant literature data to validate against; kinematic validation still gives valid relative prediction, in the sense that it differentiates between various movement speeds, various loads, etc.). In this case, the best approach is to:

   (i) Minimise the number of parameters, while keeping the model realistic;
   (ii) Measure whatever is possible, or to compare parameters to published data; and finally
   (iii) Seek the fine-tuning of remaining parameters by a sample of experts.

   At the end of this process, the third model approaches the value of the second model. This permits the use of the model as a predictive tool enabling the user to explore variations in the simulation parameters on the predicted outcome of the system it represents.

It should be noted that field trials using human participants are inherently error prone. Human performance in field conditions is influenced by a wide range of variables including population, environment, sleep patterns, and motivation, which defy tight control. In many cases this prevents the fundamental scientific requirement of repeatability for many measures. Whilst this level of error is considered to be acceptable for answering many human factors research questions, it remains a challenge for determining the basic attributes of human performance within real world environments. Once a mathematical model is able to closely approximate the complexity of the real world system and has been validated as far as possible, it offers a faster, cheaper and, importantly, a more reliable method of prediction than field-testing.

[1] "Parameter" in the physical context literally means "measured coefficient", is a constant that does not change with time. On the other hand "variables" are dynamical quantities that change with time and respond with certain outputs to certain inputs.

4.2 The Assessment of Injury Risk

To draw valid conclusions regarding the impact of given parameters on the risk of injury using traditional measurement techniques would require a proper statistical approach:

- A representative sample from the population of interest: achievable
- The establishment of standard, repeatable conditions: error prone
- The observation and measurement of actual injury: unethical
- Accurate measurement of resultant torques: not feasible.

Given that such traditional methods of biomechanical investigation are both unacceptable and unachievable the only available source of these data is within existing databases of already injured subjects. In addition to the considerable costs of this form of study, the limitations these data sources would place on the collection and statistical analysis of the data greatly reduce the scope for valid interpretation.

4.3 Predictive Modelling of Injury Risk

Currently, biomechanical modelling is unable to adequately represent the realistic complexity of the human neuro-musculo-skeletal dynamics (see section 2.2). The complexity of the model needs to match the complexity of the real situation to be really predictive. High confidence in the predictive capabilities of a model relies on it being *deductive.*

The HBE possesses the characteristics of all three of the common model types, but with much greater complexity and proximity to real human anatomy and physiology. The causal input-output relations of the general HBE formalism are qualitatively correct giving it the merit of the valid physical model. However, quantitative validation of the Simulator is faced with the two-fold difficulty:

(i) The absence of sufficiently precise measures for the empirical parameters included in the model, due largely to the limitations in suitable measures, and
(ii) The lack of consistent data within the literature for muscular torques measured in real situations. The only biomechanical technique that is currently used for estimating the forces generated by individual muscular contractions in realistic situations is portable surface electromyography, which does not give significant correlation with torques measured on fixed lab dynamometers.

This leaves *kinematic validation* (i.e., angular velocities and accelerations) with inherent imprecision of MIs as the maximum level of validation that is currently possible. This level of validation cannot give the valid prediction of absolute muscular torques and TJs (which are already inside the range prescribed by the literature), but can give the valid prediction of relative muscular torques and TJs with respect to various kinematics (i.e., speeds, accelerations, kinematic jerks) and various loads.

5. Overview of the Human Biodynamics Engine

The HBE is a sophisticated kinematics-dynamics-control model of the human *neuro-musculo-skeletal system*. The objective behind the development of the HBE is to construct a simulation of realistic[2] human biodynamics as a tool for predictive analysis and illustrative animation. The basis of the HBE is a representation of the *virtual structure and function* of the human motion system at an unprecedented level of complexity approximating the real system.

Formulated in the fashion of modern non-linear robotics, our proposed model describes the kinematics, dynamics and control of the human locomotion system, using the half-inverse dynamics approach. The full HBE engine includes over 300 active and controlled degrees of freedom. The first application of the HBE engine has been to represent the impact of loading and environmental forces on the neck and spine. The Full Spine Simulator includes 150 degrees of freedom (75 rotations plus 75 restricted translations, distributed along 25 movable spinal joints), described by sophisticated Lie-Hamiltonian kinematics, dynamics and control equations.

As in the human body, the HBE has more than two hundred rigid bones connected by joints with ligaments. All sinovial and movable spinal joints have up to three axes of constrained rotation, plus the same number of very restricted translations. All of these possible motions in all the joints gives us the total number of degrees of freedom (DOF), for both the human body and the HBE. Each DOF in the HBE has an associated *coordinate* (rotation or translation) and a corresponding *momentum*.

The human skeleton is driven by a synergistic action of more than six hundred skeletal muscles. Each of these muscles has its own excitation and contraction dynamics, in which neural action potentials (APs) are transformed into muscular forces and torques in the joints.

Excitation dynamics in the HBE is represented by a *force-time (F-t) impulse curve* (see Figure 10), having an impulse shape of AP. Contraction dynamics in the HBE is represented by a hyperbolic *force-velocity (F-v) power curve*. Both curves are said to be

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[2] This "realistic" human biodynamics is based on a two-fold mathematical philosophy of *non-linearity* and *complexity*:

1. In nature nothing is linear; therefore we can neither simulate it, nor control it, using linear engineering or statistical techniques. To understand and predict nature, we need *non-linear dynamics*; to control it, we need *non-linear control*.
2. Complexity of the model needs to reflect the complexity of the nature itself. As a measure of complexity of a system we use the number of its controlled DOF.

To deal with these two fundamental requirements, the HBE approximately matches the number of DOF of the human musculo-skeletal system, using the most general form of temporal dynamics, which includes all three mechanical categories of force: conservative, dissipative, and control.

biological invariants, i.e., they are valid on all levels of biological organization, from the muscle fibre level up to the whole body level of translation and rotation. As in the human body, muscles in the HBE are modelled by equivalent antagonistic muscle-pairs, each with their own F-t and F-v curves.

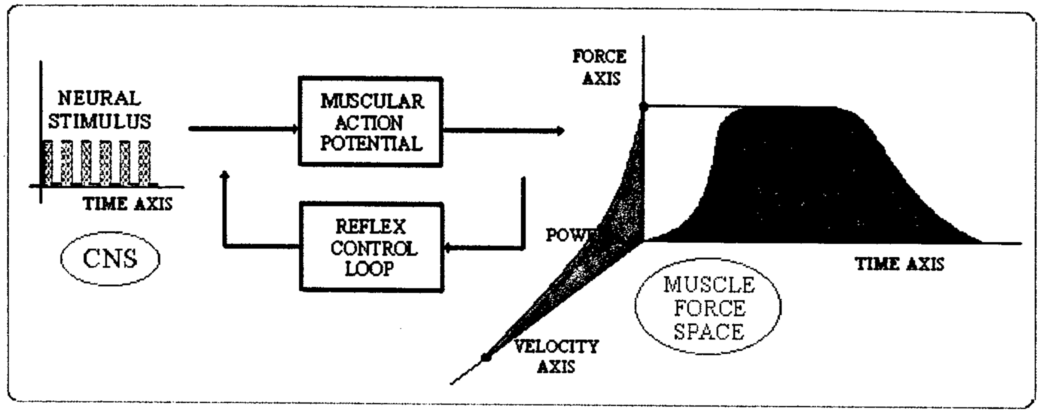

Figure 10. Muscular Excitation and Contraction Dynamics

The dynamics of complex mechanical systems, including humanoid robots, is usually described in one of the following formalisms: Newtonian, Eulerian, Lagrangian, or Hamiltonian. The HBE uses *generalized Lagrangian* and *Hamiltonian formalisms*, including *conservative, dissipative* and *driving forces*. They describe rotational joint motion, as seen in traditional human animation packages, with the important additional representation of restricted translational motion that has always been neglected, both in robotics and in human animation. The Lagrangian approach reflects engineering and variational-calculus points of view, while the Hamiltonian approach is more suitable for chaos-theory analysis, and stochastic generalizations.

The conservative part of the HBE dynamics is derived from Lagrangian and Hamiltonian functions, representing conserved mechanical energy of the system. Its dissipative part is derived from non-linear *dissipative function*, and describes non-linear joint dampings (which prevent entropy growth, otherwise present in conservative systems). Its driving part represents equivalent muscular forces and torques in all joints (or just in active joints, as used in the affine Hamiltonian control), in the form of F-t and F-v curves.

On the top of the HBE stands a hierarchical, neural-like, non-linear servo-controller. At the present stage it has two control levels:

- A lower level *force controller*, which reacts on any coordinate- and velocity-disturbances, by increasing muscular forces in the opposite directions. It resembles an *autogenetic reflex motor servo*[3], acting on the spinal level of the human motor control.

- A higher level *velocity controller*, which performs the self-stabilizing and adaptive tracking of desired motion trajectories with the previously defined musculo-skeletal dynamics. It resembles the self-organizing, associative function of the human *cerebellum*. It is designed using a high-order Lie-derivative formalism.

The top level, cortical control, is planned for the future development of the HBE. Its objective will be to *define the desired trajectories* for all movable joints for each desired movement (e.g. to translate the command "sprint forward and jump" into optimal joint trajectories). For this, we intend to use a form of adaptive topological fuzzy logic (a hyper joystick). However, regarding the virtually infinite number of possible desired movements and their variations, this would require a huge knowledge base of movements.

Joint coordinates and momenta (both rotational and translational) are HBE *system variables*. All the bone parameters are derived from the user's body *weight* and *height* using standard anthropometrics tables. All the *driving parameters*, included in F-t and F-v curves, are derived from the user's *general strength, speed* and *reaction time*. The *HBE-simulation* represents the evolution, over time, of the system variables, based on the user-defined body and load parameters.

All kinematics, dynamics and control equations of the HBE system have been first modelled, using generalized Lie-Hamiltonian formalism, inside the computer algebra system *Mathematica*™. Symbolically pre-processed equations have been subsequently implemented in *Delphi*™ compiler for MS Windows™, and integrated using our own symplectic-matrix integrator, to make a standalone Full Spine Simulator.

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[3] Voluntary contraction force F of human skeletal muscle is reflexly excited (positive feedback F^{-1}) by responses of its *spindle receptors* to stretch and is reflexly inhibited (negative feedback $-F^{-1}$) by responses of its Golgi tendon organs to contraction. Stretch and unloading reflexes are mediated by combined actions of several autogenetic neural pathways, forming the so-called 'motor-servo'. The term 'autogenetic' means that the stimulus excites receptors located in the same muscle that is the target of the reflex response. The most important of these muscle receptors are the primary and secondary endings in muscle-spindles, sensitive to length change - positive length feedback $+F^{-1}$, and the Golgi tendon organs, sensitive to contractile force - negative force feedback $-F^{-1}$.

6. Future Directions

For the sake of safety and performance it is important to take account of a user's biodynamic capabilities when considering the selection of systems that must be worn or carried. As this report has illustrated, basic principles of load reduction and balance around the human musculo-skeletal system should be addressed in the design and selection of any carried equipment. At the same time the true impact of these loads on the human within a dynamic environment has yet to be fully determined. There remains disagreement in the field of medicine regarding the causes of injury and the treatment of pain and dysfunction within the musculo-skeletal system. The development of the HBE has opened new opportunities for examining the issues of loading and injury in greater detail than previously possible. It forms the basis of what promises to be a powerful tool for the evaluation of loading within a dynamic environment on human safety and performance.

A current version of HBE-simulator through its representation of the full human spine is the first step in the implementation of the complete, movable, human skeleton with 200+ bones. It provides a simulation of the general motion of the spine approaching the complexity of the real system. As a full spine model it is unprecedented in its complexity, including all possible forces, several hundred degrees of freedom, physiological-like control system (Figures A1 and A3 show full spinal stabilization from any initial spinal position) and variety of different movement/force analyses for all joints. In doing so it is distinctly different from other current models in which the representation is either much simpler or limited to a translational representation.

Continued development of the model will take a number of directions. Firstly, it will seek to further refine and calibrate the model against available sources to improve its predictive power. A start has been made to extending the simulation to include the representation of the limbs, including their dynamics and control. This will open up enhancements to the model's functionality including the investigation of a range of real world movements including walking, running, jumping, lifting and shooting. The existing modelling of stresses from the external environment will also be extended and refined where relevant to a range of areas, e.g. vehicle motion, car crash, parachute landing, aircraft ejection, and explosion. As the HBE evolves it has the potential to resolve many questions surrounding the physical safety and performance of the military service personnel across a wide range of dynamic operational environments.

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Appendix A: On the mechanism of NVG induced injury in the military context

A.1. HBE Simulation Outputs: Single Stabilising Movement With and Without NVGs

Here we present simulation charts on the spinal data derived from a single stabilising movement with and without an NVG-like HMD, with a total mass of 3.5 kg. The spine has 25 movable joints, each with three restricted rotations (dominant) and three translations (highly restricted).

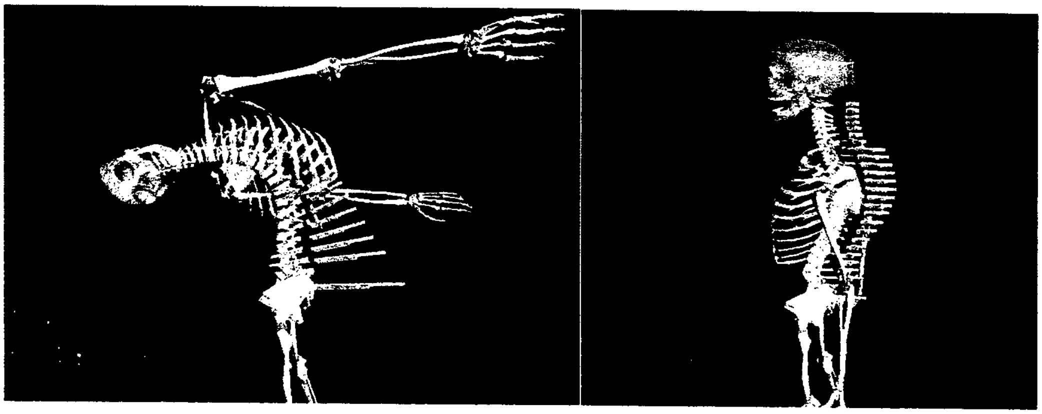

Fig A.1. Simulation start-left and simulation end-right

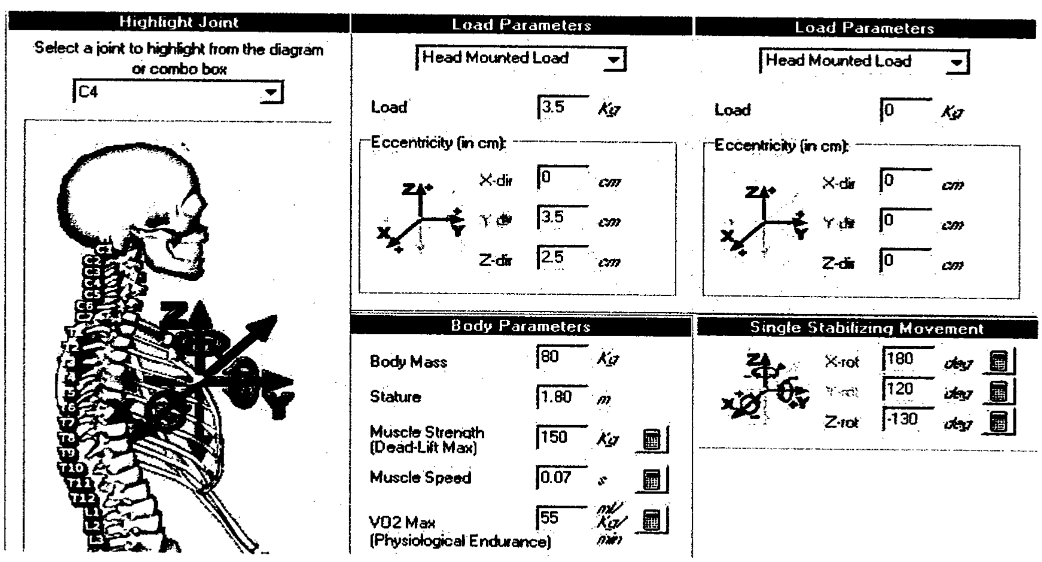

Fig A.2. Simulation data: with and without NVGs

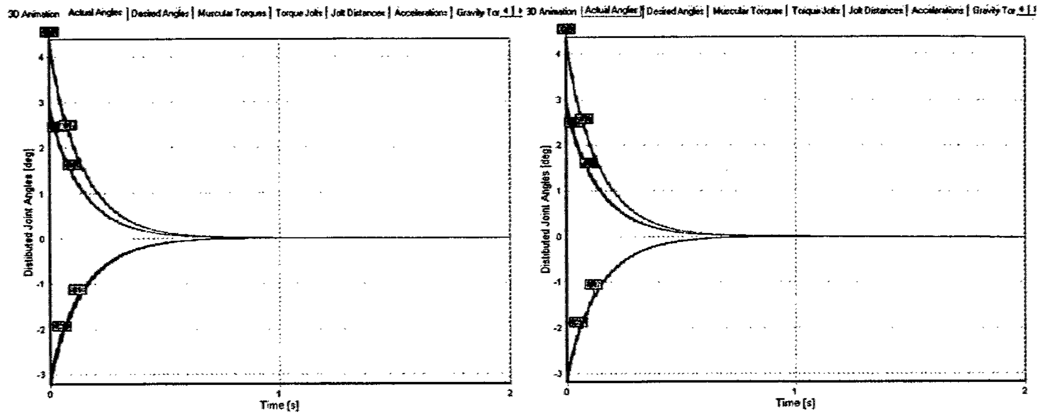

Fig A.3. Simulated joint angles, with and without NVGs

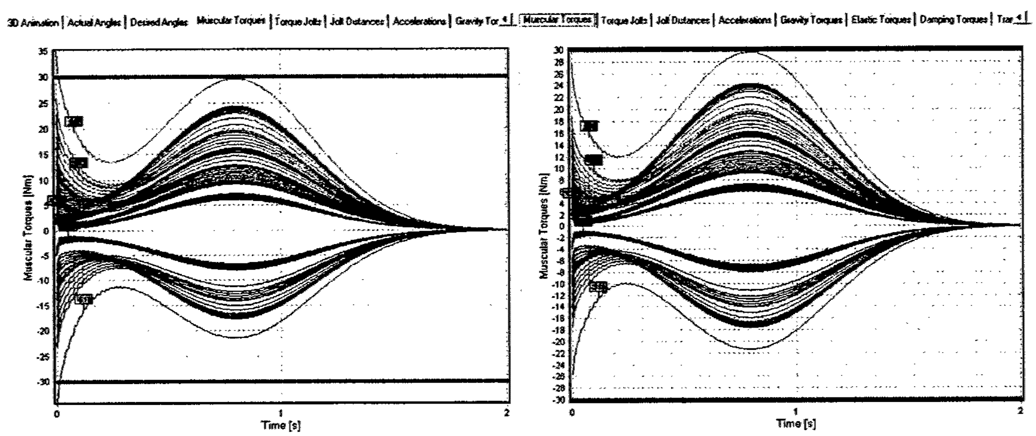

Fig A.4. Simulated joint torques, with and without NVGs

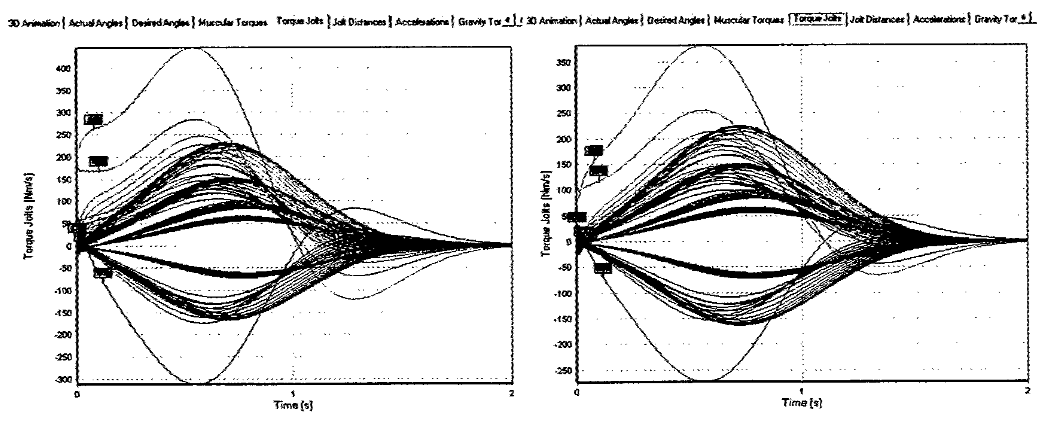

Fig A.5. Simulated joint torque-jolts, with and without NVGs

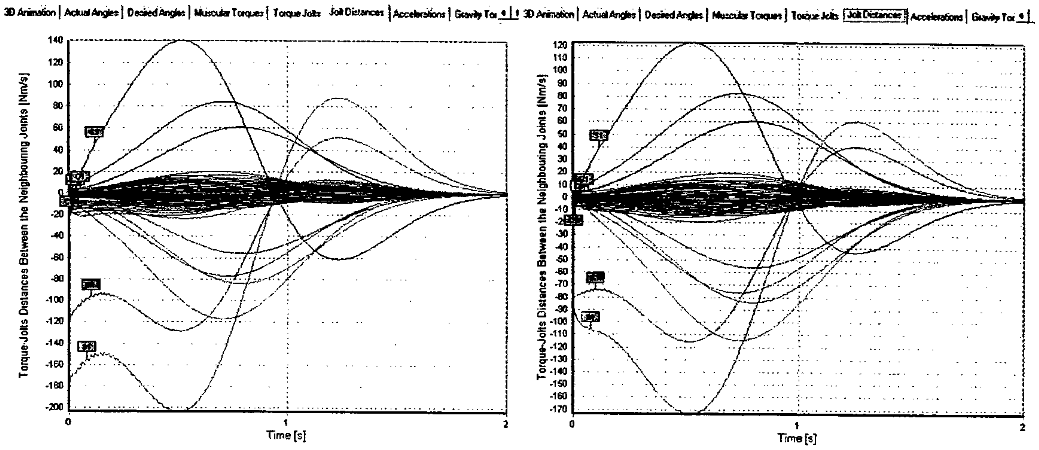

Fig A.6. Simulated jolt-distances along the spine, with and without NVGs

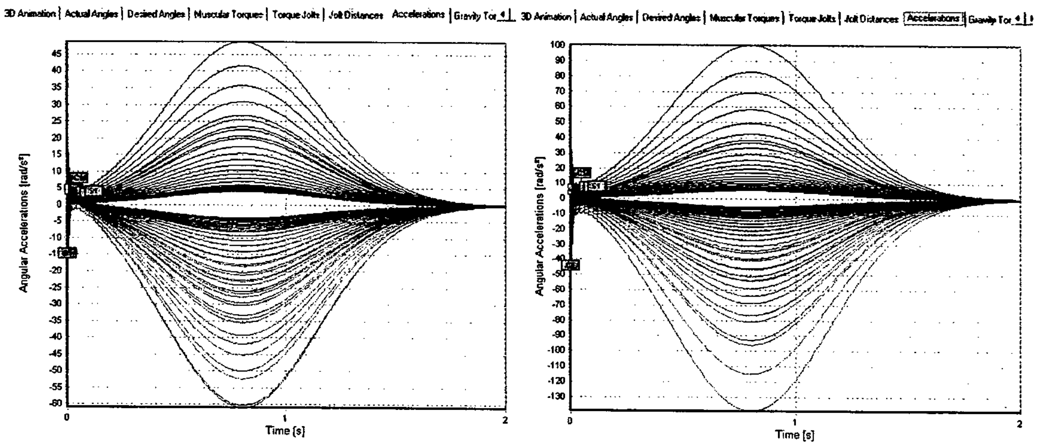

Fig A.7. Simulated joint angular accelerations, with and without NVGs

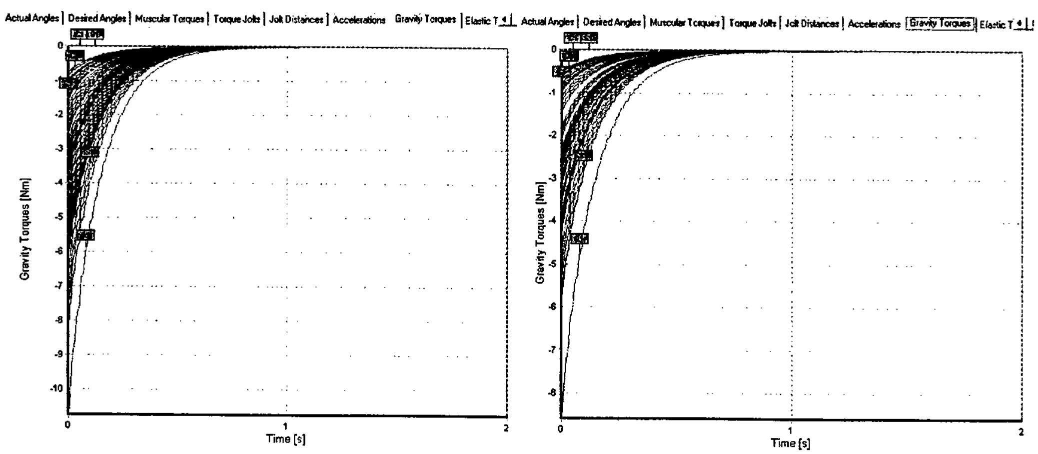

Fig A.8. Simulated joint gravity torques, with and without NVGs

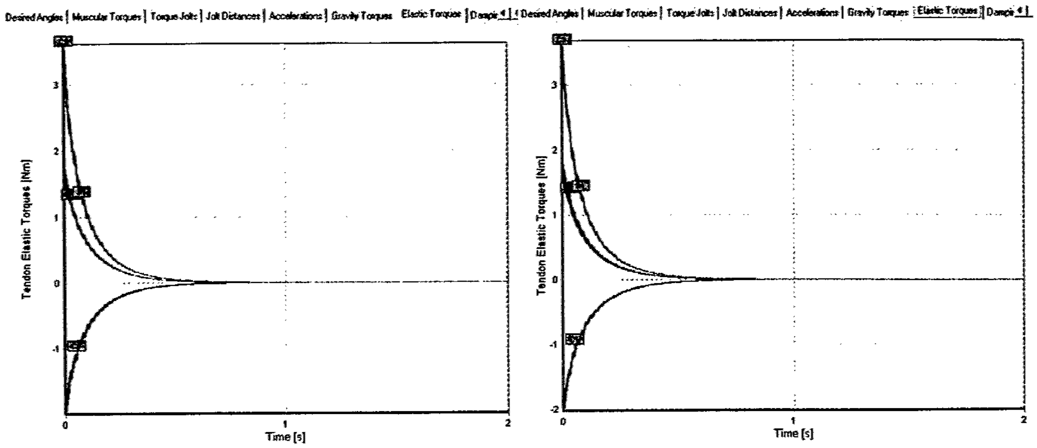

Fig A.9. Simulated joint discus elasticity torques, with and without NVGs

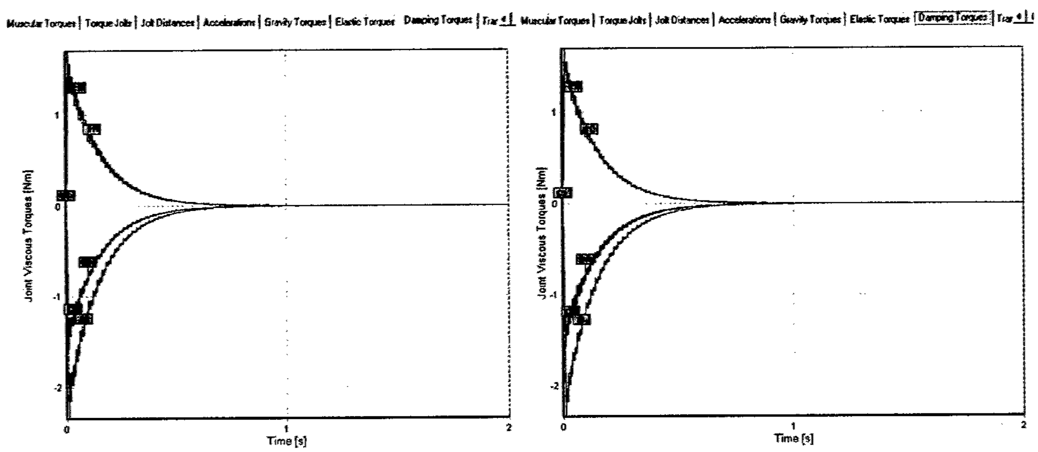

Fig A.10. Simulated joint discus damping torques, with and without NVGs

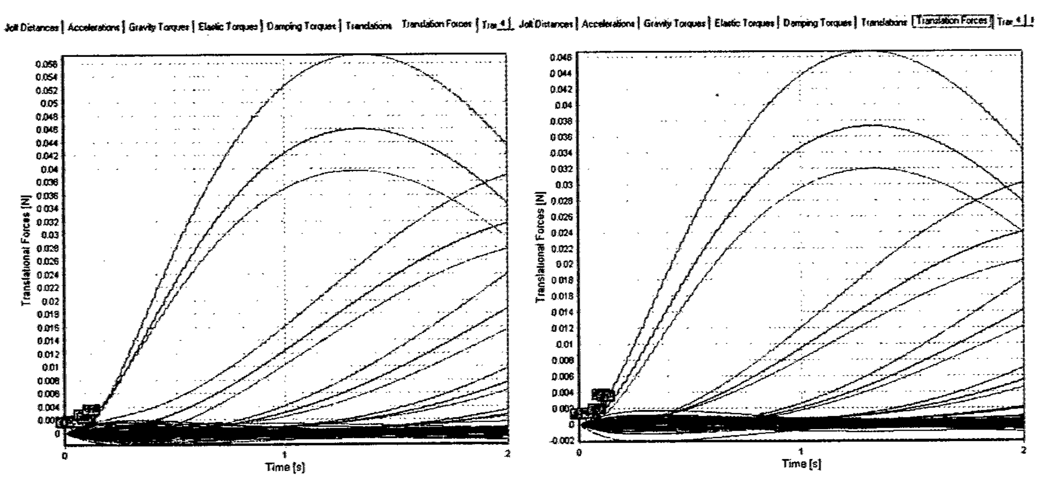

Fig A.11. Simulated joint translational forces, with and without NVGs

A.2. New Insights Into the Mechanism of Spinal Injuries, Based on the HBE-Calculated Torque-Jolts

In the literature on musculo-skeletal injuries, the description and classification of spinal injuries is usually based on the *principal loading* applied to the spine (see [24], [25]). The principal loading hypothesis predicts musculo-skeletal injury resulting from a combination of *tension, compression, bending* and *shear* that occurs as a direct consequence of the application of force beyond the joint thresholds in these dimensions. In this comment we offer an alternative view regarding the mechanisms underlying functional and structural musculo-skeletal injury, and the conditions under which injury might occur.

To be recurrent, musculo-skeletal injury must be associated with a histological change, i.e. the modification of associated tissues within the body. However, incidences of *functional* musculo-skeletal injury, e.g. lower back pain, generally show little evidence of *structural* damage [26]. The incidence of injury is likely to be a continuum ranging from little or no evidence of structural damage through to the observable damage of muscles, joints or bones. The changes underlying functional injuries are likely to consist of torn muscle fibres, stretched ligaments, subtle erosion of join tissues, and/or the application of pressure to nerves, all amounting to a disruption of function to varying degrees and a tendency toward spasm.

Given the assumption that an injury is the effect of tissue damage, it becomes important to understand the mechanism for such damage. The damage to the component tissues that make up the body's structural system is a consequence of a failure of any part of that system to dissipate externally imposed energy to either adjoining structural components, or to the surrounding environment. The components of the spine will break at the point that all of its vertebrae are at the extreme of their rotation and translation and when the tissues at their interface are at the limits of their elasticity. At this point, where no further motion is possible, the addition of further force will result in the failure of the system in the form of tissue damage at the weakest point.

At the same time, there is little evidence that static force, either internal (muscular), or external (loading), is the prime cause of injury, though it may aggravate an old injury. For example, maximal isometric muscle contractions, even in prolonged strength-endurance tests, are commonly used in the training of a variety of sports without injury. Similarly, the stretching techniques of gymnasts and ballet dancers are considered to be safe when performed very slowly i.e., under quasi-static conditions. These examples help to illustrate the body's ability to endure extreme static forces that, based on the theory of principal loading, would be expected to induce injury. Experience suggests that injuries occur, primarily, under situations where control over the situation is lost e.g., an abrupt change of direction when running or a shift in balance when lifting an awkward weight. We are not injured under high loading semi-static conditions because we are able to position our bodies (joints) in such a way that we dissipate forces through compression rather than by potentially damaging motions of twisting or shearing. When control is lost, the joints may be forced in a rotation or direction for which they are not adapted to support.

Whilst it is evident that the body can endure extreme forces under static conditions, injuries appear to result from conditions where much lower peak forces are reached, if the transfer of energy is rapid and localised. In other words the primary cause of both functional and structural injury is the *local dynamism* of the force.

For example, in a review of experimental studies on the role of mechanical stresses in the genesis of disk degeneration and herniation [27] the authors dismissed simple mechanical stimulations of functional vertebra as a cause of disk herniation, concluding instead that a complex mechanical stimulation combining forward and lateral bending of the spine followed by violent compression is needed to produce posterior herniation of the disk. Considering the use of models to estimate the risk of injury in [28] the authors emphasise the need to understand this complex interaction between the mechanical forces and the living body. The role of this combination of factors is underlined in [29] in which the authors indicated that compressive and shear loading increased significantly with exertion load, lifting velocity, and trunk asymmetry. In [30] the authors analysed the influence of dynamic factors on triaxial net muscular moments at the L5/S1 joint during asymmetrical lifting and lowering. They concluded that their results demonstrated that dynamic factors do influence the load on the spine for both lifting and lowering.

We define the measure of local dynamism as the *"jolt"*, which is the time rate of change of the force. In other words, we suppose that the force is not constant i.e., static, but rather that it is changing, and that we can measure the rate of its change (see [31]). As the rate of change of the force increases it tends towards a "unit impulse[4] function" or Dirac's "delta function." It is this instantaneous application of force that we consider to be the cause of injuries, at first only functional, i.e., reduction in function without obvious damage to tissue, and later even structural, i.e., loss of function directly attributable to specific localised tissue damage, e.g. a herniated intervertebral disc. It is the instantaneous nature of this applied force that defies the tissues' ability to dissipate energy with the consequence of raising the energy at that point to a level beyond the tolerance of the structural bonds that maintain the integrity of the tissue. The higher the rate of force change, the higher the risk of injury. Therefore, the same energy, or the same mechanical impulse, applied to the human body within the normal anatomical range of motion, can cause two totally different outcomes: the one with a constant or very slowly changing force would not cause any damage, while the other which is rapidly changing would cause injury (see Figure A12).

[4] Here, mathematical "unit impulse function" should not be confused with the "mechanical impulse," $\int_0^t F(t)dt$, which is the total area under the force-time curve.

Besides the time-locality of the force impact, we need to consider the space-locality measured as a pressure, i.e., the force per body area. In the spinal context this means the local point at which we expect an injury. When evaluating the risk of injury, it is not sufficient to simply consider a single translational force applied to the whole spine, neglecting the location of the load. The combination of additional mass and its eccentricity from the normal posture of the spine, measured by the distributed moments of inertia, i.e. the mass distribution, must be considered.

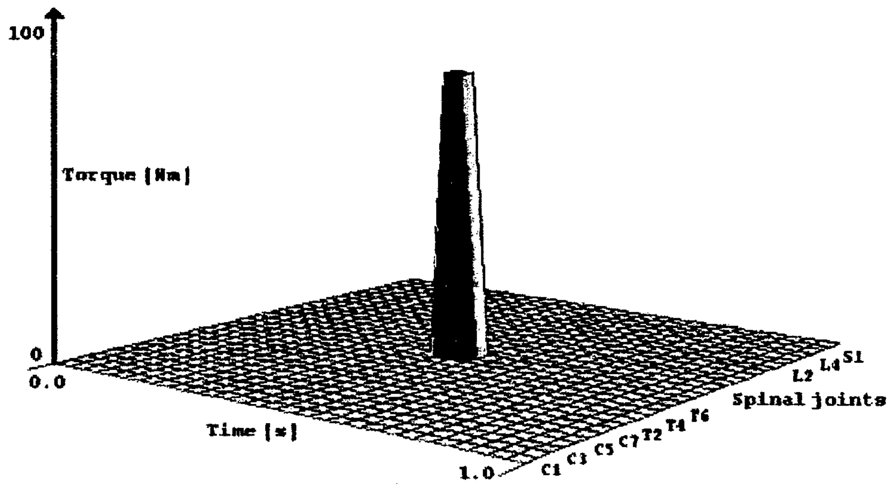

Fig A.12. Space-time Delta-Torque as a main cause of spinal injuries

When determining a realistic mechanism of human spine injury, it must be emphasized that the natural dynamics of the spine is predominantly rotational, not translational as presumed by the principal loading hypothesis. Translation is evident in many cases of real world spinal injury, e.g. spinal compression. In the case of relatively low forces, e.g. carrying a weight on the head, the symmetry of the body tends to allow these compressive forces to be absorbed without injury. In cases where much a larger compressive force is applied in a short time interval e.g. during an accident, the ability of the body to maintain symmetry is lost. Slight asymmetry in the body will have the effect of converting the external translational force into a rotational motion distributed along the structural components of the skeleton. At the extremes of a single joint's rotation, where no further movement is available to dissipate the force, disk deformation in the form of herniation will occur. At the global level, the importance of rotational dynamics can be illustrated by the accepted work practice for lifting and handling, summarised by the maxim: "Bend the knees to save the back". By maintaining a "straight" spine with the weight close to the body when lifting we reduce the rotational forces on the spine thus reducing the risk of injury. Mechanically, this means that we need to consider the action of various torques T on the human spine, rather than translational forces F (loadings). Torque T acting in the (spinal) joint is the (vector) product of the force F and its lever arm r for the particular joint. So, even with the constant loading *F*, the torque at the particular (spinal) joint can change rapidly due to the change of a particular lever arm *r*, and this change might cause an injury. That is why, in our opinion, the best measure for the risk of (spinal) injuries is the *time rate of change of the total joint torque*, i.e., the *torque-jolt*, technically *dT/dt*, measured in *Nm/s*. The higher the torque-jolt *dT/dt* at the certain point along the spine, the higher the risk of injuries, at first only functional, progressing towards later structural injuries.

It should be noted that the term *torque-jolt* does not appear in the dynamic literature. There is a *kinematic* (non-material) quantity called *"jerk,"* which denotes the rate of change of *translational* acceleration. What we propose here is both an *inertial* (material) and a *rotational* form of a *jerk*, which we call the torque-jolt *dT/dt*, with three components around each of the three world axes (X, Y and Z) that are each time dependant and distributed along the spine, i.e., localised at each movable spinal joint:

Torque-jolt = Kinematic jerk × Joint lever arm × Inertia moment

Therefore, our proposed new measure of the risk of spinal injury, Torque-Jolt (see Figures A5-A6 for its implementation in the full spine HBE simulator), includes the combination of space-time localised rapid movement, arising from either internal or external sources, and the distribution of additional masses on the body [32,33].

We suggest that the energy imparted by torque-jolt is a root cause of injury. At low levels, torque jolt is dissipated within the associated tissues without lasting effect. Above various thresholds it will modify various tissues, i.e. cause damage, initially at a very low level and essentially invisible to diagnostic techniques. As the magnitude of the torque jolt increases, the level of tissue damage would tend to increase up to a point where obvious structural damage may be observed. We would suggest that a lasting modification to localised tissues (muscles, nerves, tendons, disks) is required for functional disturbance to be both recurrent and localised. This is not to say that all tissues exhibiting this disturbance have suffered damage.

If we take the example of muscle spasm, this may be due to chemical imbalances, but usually follows an injury. After about two weeks, if the basic injury has failed to repair, the spasm maintains an unnatural muscle contraction leading to the formation of knots. This happens most frequently for the small deep muscles of the spine, the rotatores, multifidus, interspinales, and intertransversarii. The stronger erector spine group, the iliocostalis, longissimus and spinalis, tends to be the last to be injured. Nevertheless, it is ready to go into painful spasm, whenever any of the smaller muscles is injured. The underlying point is that whilst this larger muscle group exhibits pain and a tendency to spasm without evidence of injury, the trigger for this pain and spasm is an underlying tissue change responsible for changes in the electrophysiology of surrounding tissues. In an attempt to reduce the risk of developing the symptoms of functional disturbance we are currently of the view that we must predict this localized tissue change. Our proposed mechanism and method of prediction is the torque jolt.

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V Ivancevic and N Beagley

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19. ABSTRACT
This report reviews ergonomic and biodynamic factors related to the introduction of head mounted loads for military land environments. It presents an overview of current knowledge pertaining to the assessment of head mounted load limits. Limitations in the predictive ability of existing modelling and measurement techniques are discussed. The report introduces a novel biodynamic model, currently under development by DSTO, designed to overcome these limitations. LOD's Human Biodynamics Engine (HBE) is a realistic model of the human body motion within a dynamic environment. The model is unprecedented in its representation of the full, distributed, three-dimensional neck & spine dynamics and in its incorporation of reflexive and cerebellar-like control of spinal motion. The model is used to investigate specific considerations of head mounted load design and their use in an operational context.

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