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A description of spinal fatigue strength

Gerd Huber [a,*], Katrin Nagel [a], Daniel M. Skrzypiec [b], Anke Klein [c], Klaus Püschel [c], Michael M. Morlock [a]

[a] Institute of Biomechanics, TUHH Hamburg University of Technology, Germany
[b] Leeds Institute of Rheumatic and Musculoskeletal Medicine, University of Leeds, Leeds, UK
[c] Dept. of Legal Medicine, University Medical Centre Hamburg-Eppendorf, Germany

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ABSTRACT

Understanding fatigue failure of the spine is important to establish dynamic loading limits for occupational health and safety. In this study experimental data were combined with published data to develop a description of the predictive parameters for spinal fatigue failure. 41 lumbar functional spinal units (FSUs) from cadaveric spines (age 49.0 ± 11.9 yr) where cyclically loaded. Three different levels of sinusoidal axial compression (0–3 kN, 0–2 kN or 1–3 kN) were applied for 300,000 cycles. Further, published data consisted of 70 thoracic and lumbar FSUs loaded in axial compression for 5000 cycles. Cyclic forces ranged from lower peaks (F_{min}) of 0.7–1 kN to upper peaks (F_{max}) of 1.2–7.1 kN. Based on Wöhler analysis, a fatigue model was developed accounting for three parameters: I) specimen-specific scaling based on the endplate area, II) specimen-specific strength dependency on age or bone mineral density, III) load-specific correction factors based on F_{max} and F_{min}.

The most predictive model was achieved for a combination of F_{max}, endplate area and bone mineral density; this model explained 61% of variation ($p < 0.001$). A model including F_{max}, endplate area and age explained only 28% of variation ($p < 0.001$). Inclusion of a load-specific correction factor did not significantly improve model prediction of fatigue failure.

This analysis presents the basis for the prediction of specimen-specific fatigue failure of the lumbar spine, provided the endplate area and bone mineral density can be derived.

1. Introduction

The systematic approach for fatigue testing, introduced by August Wöhler in 1860, dramatically improved engineering of cyclically loaded systems. The empirical curves enable the prediction of the maximum strength for a given number of consecutive loading cycles.

The basic Wöhler approximation disregards extreme cases for either a few cycles of high load magnitude (close to the ultimate-strength), for low load magnitudes that could be applied indefinitely. Finite life fatigue strengths can be characterised by a pragmatic equation, describing the bearable stress (σ) by a power function based on the numbers of cycles (N), an exponent (κ) and two scaling constants ($\sigma_{\text{infinite life}}$, N_0), with both sides of the equation dimensionless:

$$\frac{\sigma}{\sigma_{\text{infinite life}}} = \left(\frac{N}{N_0}\right)^{\kappa} \quad (1)$$

If 0 logarithmic scales are used on both axes, the description of these so called *S-N*-curves ('stresses' vs. 'number of cycles') turns out to be linear (Basquin, 1910). Experimental data can thus be used to determine the slope (κ) and the combined constants.

$$\log_{10}(\sigma) = \kappa \cdot \log_{10}(N) + \left(\log_{10}(\sigma_{\text{infinite life}}) - \kappa \cdot \log_{10}(N_0)\right) \quad (2)$$

The slope will be negative, since higher cycles to failure will decrease the failure stress. These graphs are based on a large number of individual measurements for a specific material or a specific structure and also for standardized loading – described by waveform, frequency and the applied force peaks (minimum and maximum, respectively). The relation of the two force peaks determines whether pure compression, pure tension or alternating loads are applied. Furthermore, loading with higher amplitudes might decrease the cycles to failure compared to loads with smaller amplitudes but the same maximum force.

For the highly variable skeletal system such data is scarce. However, occupational health requires evidence-based guidance and limits to guarantee a safe working environment. Whole body vibrations (WBV) pose a particular problem because even

* Correspondence to: Hamburg University of Technology, Biomechanics, Denickestraße 15 (K), 21073 Hamburg, Germany. Tel.: +49 40 428 78 31 53; fax: +49 40 428 78 29 96.
E-mail address: g.huber@tuhh.de (G. Huber).

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relatively small loads experienced over long period of time might cause structural failure.

Measurements of human vibration transmissibility, using sensors screwed on the spinous process (Panjabi et al., 1986, Pope et al., 1998), revealed that vibrational loads are transferred over the spine, but the measurements were limited by the external position of the sensors and also by their short term nature, due to ethical restrictions-fatigue experiments are not possible. Besides the ethical and practical considerations regarding in-vivo measurements of mechanical properties of internal structures, the inter-individual variabilities pose a challenge to obtaining a consistent set of data.

Numerical modelling has been used to estimate skeletal loading during exposure to external vibration. These approaches have a limited capacity to predict fatigue fracture. They are restricted to the analysis of structural stiffness and the prediction of internal accelerations (Bass et al., 2005) or forces (Seidel et al., 1998, 2008), which can only be related to the ultimate-strength of substructures measured under specific load conditions in-vitro or to limits framed by legal regulations (ISO 2631 Part 5). However, stiffness and ultimate-strength are commonly not the parameters of interest – the primary stability of implants or the evolution of tissue or bone degeneration are mostly related to cyclic loading. Such loading arises due to human locomotion, with a comparably small numbers of cycles, or by high frequency cyclic loading caused by machines, such as construction or farming vehicles. The health risk caused by WBV in the workplace has been under considerable debate and society would benefit from a greater volume of fatigue strength data for spinal structures.

Since in-vivo experiments are precluded in-vitro data might bridge this gap for passive structures such as the spine, notwithstanding that fatigue experiments are time consuming and pose the challenge of creating a suitable and meaningful testing environment for cadaveric tissue over prolonged periods. In the late 1950s, multi-segmental specimens were used for fatigue analyses, which provide less specific failure information, and specimens were embalmed (Hardy et al., 1958), which is known to modify the mechanical properties (Wilke et al., 1996). Investigations performed at room temperature (Hansson et al., 1987; Liu et al., 1983) have limited utility, since ovine FSUs immersed in water at 37 °C exhibited less stiffness than those at room temperature (Costi et al., 2002) and ovine bone samples warmed to 37 °C survived double the number of cycles to failure at the same stress (Carter and Hayes, 1976). Due to these limitations an adequate conjunction of fatigue measurements from different specimens is not yet available. In addition, a concept to account for differences among spinal levels and subjects is required.

This study aims to provide a description of the spinal in-vitro axial compression fatigue behaviour.

2. Methods

Functional spinal units (FSUs) from human donors were used for in-vitro experiments. A cohort of specimens was exposed to high numbers of compression fatigue cycles (high-cycle cohort). Results were pooled with another cohort from Brinckmann et al. (1988) that were exposed to similar experimental conditions, but sustained moderate numbers of compression fatigue cycles (moderate-cycle cohort).

2.1. In-vitro experiment

The high-cycle cohort comprised of in-vitro experiments by the authors. The FSUs ($n=41$) were taken from the lumbar spine of human donors (age 49.0 ± 11.9 yr)-either L2/L3 ($n=15$) or L4/L5 ($n=26$). They were wrapped in wet gauze, sealed in plastic bags and stored deep-frozen (< -20 °C). This storage mode is accepted for its negligible effect on the spinal creep behaviour (Dhillon et al., 2001) and stiffness (Gleizes et al., 1998). The frozen spines underwent computed tomography (CT) scanning (S5VA40A, Siemens, Germany & Mx8000 IDT 16, Philips Healthcare, The Netherlands). Pathological specimens were excluded. The specimens were thawed at room temperature overnight. Muscles were removed, while adjacent ligaments were left intact. Both vertebrae of the FSU were potted in metal holders (RenCast FC53, Switzerland) maintaining the transverse mid-plane of the intervertebral disc parallel to the flanges and consequently perpendicular to the loading actuator above. A six degree of freedom load cell (30031, Huppert, Germany) was positioned below the specimens. After preconditioning and non-destructive parameter measurements (~90 min, maximal axial force 2.0 kN, maximal anterior–posterior force 0.4 kN) the FSUs were exposed to sinusoidal axial compression (300,000 cycles, 5 Hz). Consequently, the testing duration amounted to ~18 h. Anterior–posterior shear load was maintained at 0 kN under force-control, while the four remaining degrees of freedom (lateral shear, flexion–extension, axial rotation and lateral bending) were constrained. Three different levels of axial compression were applied. Peaks of repeated sinusoidal loading ranged from 0–3.0 kN ($n=14$), 0–2.0 kN ($n=20$) and 1.0–3.0 kN ($n=7$). The peak force of 3.0 kN equates to approximately 40% of the estimated compressive ultimate-strength (Brinckmann et al., 1989). Under comparable conditions axial compression of 2.0 kN induces pressures of approximately 1.4 MPa in the nucleus pulposus (Huber et al., 2005)which is similar to the one measured during lifting 20 kg in-vivo (Wilke et al., 1999). The loading scenario can be considered to represent the high physiological range, even though cyclic axial loading with those frequencies rather represents loading due to environmental vibrations in a working scenario with heavy machines, helicopters, or speedboats rather than active lifting scenarios.

During loading the height of the FSU was recorded. The axial loading caused a characteristic compressive creep curve, because of water loss and viscoelastic deformations of the disc. In the log space this creep appears almost linear. Failure of the FSU, and the corresponding number of cycles to failure, were determined in accordance to former publications (Brinckmann et al., 1988, Huber et al., 2010b) by the onset of unsteadiness – step or sudden height decrease – in the creep curve (Fig. 1) by two independent researchers (averaged result, repetition of cases with a deviation above 10%). These failure events manifest in severe breakdown of the mechanical behaviour of the FSUs. Post-test examinations were performed to expose the biological structure that might have caused those changes.

During testing the specimens were immersed in saline solution at 37 °C. To minimise degeneration of the specimen, 10 ml/l Penicillin/Streptomycin (PAA Laboratories GmbH, Austria) was added. Further descriptions of the test-setup can be found elsewhere (Huber et al., 2010a, Nagel et al., 2013).

Data from the above experiments was combined with the cohort from Brinckmann et al. (1988) that sustained moderate numbers of loading cycles. These FSUs ($n=70$) were taken from the lower thoracic and the lumbar spine levels of human donors (age 51.6 ± 17.7 yr). The spinal levels were: L1/L2 ($n=4$), L2/L3 ($n=14$), L3/L4 ($n=6$), L4/L5 ($n=16$), T12/L1 ($n=16$), T10/T11 ($n=6$) or T11/T12 ($n=8$). The specimens underwent CT scanning in an environment that was simulating the spine within the body. This body phantom consisted of a PVC container filled with water. The specimens were mounted in metal cups using bone cement, just covering the outer endplates. The specimens were exposed to a stream of air of 100% humidity heated to 36.5 °C for 30 min, which continued during testing. After preconditioning (~15 min, axial force 0.7–1.0 kN, maximal anterior-posterior force 0.4 kN) the FSUs were exposed to axial compression with a triangular waveform (5000 cycles, 0.25 Hz). The duration of the measurements was about 5:33 h. The loads were applied via a cylinder with a flat face, restricting flexion-extension and lateral bending. The peak force of the applied axial cyclic compression was specified based on the ultimate-strength of adjacent FSUs. Depending on the FSU, 20–70% of the predicted ultimate-strength was applied. The lower force peaks started at 0.7–1.0 kN and the higher force peak (maximal compression) was between 1.2 kN and 7.1 kN.

2.2. Wöhler equation

A suitable Wöhler description for spinal specimens should account for three issues: (I) specimen-specific scaling, (II) specimen-specific strength expression, (III) load-specific correction factor.

Specimen-specific scaling (I) could be achieved by a specimen-characteristic effective area ($A_{\text{effective}}$) in the form of the relation between load (F) and stress (σ) – especially if axial loading is regarded. It is unclear which cross sectional area (disc, endplate or vertebra) would fit best, but it might be appropriate to at least assume a linear relation with the endplate area (A) of the FSUs.

$$\sigma = \frac{F}{A_{\text{effective}}} \tag{3}$$

$$A_{\text{effective}} = \frac{A}{\bar{A}} \cdot c_A \tag{4}$$

The constant c_A accounts for the linear relation between the chosen endplate area (A) and the unknown effective reference area. $\bar{A}$ represents a normalising factor and will be set to the representative size of a typical endplate. This will keep this part of the equation dimensionless. In the following such normalisations will be also used for other parameters.