Principles of Spinal Biomechanical Testing

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Introduction

The spine is a complex mechanical structure. Its purpose is to protect the spinal cord and nerve roots and transmit the weight of the upper body to the pelvis while providing a high degree of flexibility to the body. Spinal biomechanics is the study of how forces interact with the spine. These forces can be internal (e.g. imposed by muscles) or external (e.g.from trauma, including surgical procedure). An understanding of spinal biomechanics is essential for understanding most areas of spinal surgery, especially spinal trauma and spinal instrumentation. Additionally, advancements in spinal procedures and novel treatment technologies have been generated from biomechanical studies. This article will break down spinal biomechanics and its applications in two sections: the first section will cover fundamental knowledge of biomechanics, and the second section will cover basic types of biomechanical testing.

Fundamental concepts and anatomy

The spine consists of discrete bony elements (vertebrae) connected by ligaments, kept separated by intervertebral discs and articulating joints, and moved by muscular activations that are simultaneously controlled by the nervous system. Spinal stability is maintained by these three mechanisms: 1. The musculoskeletal system (active system), 2. The spinal column (passive system), and 3. The nervous system (controlling the active system)[1]. Under physiologic conditions, these three systems maintain mechanical stability while the spinal column is moving within a normal range of motion. A three-dimensional Cartesian coordinate system can be used to demonstrate how spinal segments move with respect to three axes (X, Y, and Z). This provides six potential spinal movements. (table1). The spine is designed to physiologically rotate around the three axes (X, Y, and Z). However, translation along any axis is typically a non-physiologic movement[2].

The complex interrelationships of the spine can be simplified by dividing the spinal column into small units known as functional spinal units (FSU). An FSU is comprised of a superior vertebra, an intervertebral disc, an inferior vertebra, and an osteoligamentous complex. Excluding the upper cervical spine (C1, C2), a FSU is connected by 10 ligaments (Fig. 1), including the anterior longitudinal ligament (ALL), the posterior

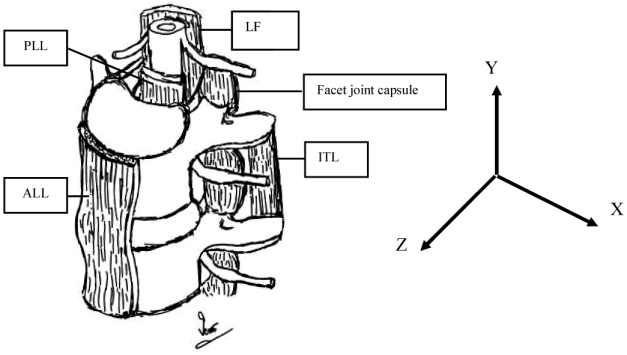

Fig. 1 (Left) a functional spinal unit (FSU), ALL; anterior longitudinal ligament.
PLL; posterior longitudinal ligament, LF; ligamentum flavum, ITL; intertransverse ligament. (Right) Three-dimensional axes of the spine.

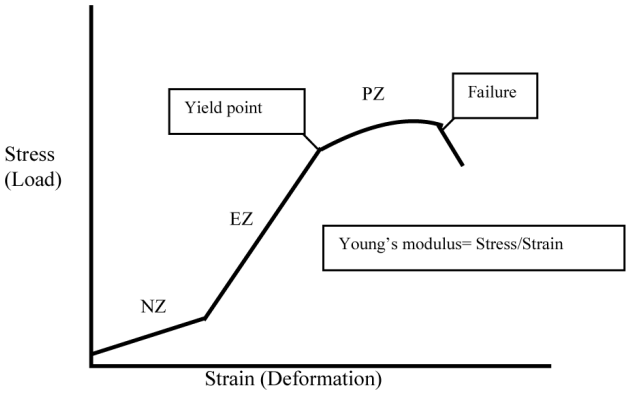

Fig. 2 A load-deformation curve (Stress/Strain curve) for a biological tissue such as a ligament.
NZ; neutral zone, EZ; elastic zone, PZ; plastic zone (permanent deformation), Range of motion (ROM)= NZ+EZ. (Adapted from White AA, Panjabi MM: Clinical biomechanics of the Spine, 2nd ed. Philadelphia, Lippincott Williams & Wilkins, 1990, p 21)

longitudinal ligament (PLL), the Ligamentum flavum (LF), the interspinous ligament, the supraspinous ligament, both sides of the intertransverse ligament (ITL) (connect between the transverse process), and the capsular ligaments (CL) of each facet joint. These ligaments serve to protect neural structures by restricting the motion of each an FSU. The Ligamentum flavum is the only spinal ligament which is primarily composed of elastin, while the others are primarily collagenous.

Mechanically, spinal ligaments are viscoelastic with non-linear elastic components. There are three zones of the Load-deformation curve (Figure 2) which correspond to different phenomena, ultimately leading to tissue failure. If a load (moment or force) is applied to an FSU, the unit first is displaced from a neutral position to a position where resistance is encountered. The initial lax region of the load-deformation curve is termed the neutral zone (NZ) or lax zone. Within the NZ, the spine can undergo relatively large motions while requiring little muscular effort. The next zone is a stiffer region, termed the elastic zone (EZ). The size of the elastic zone depends on the elastic modulus of each specific tissue. Within the FSU, elasticity is greater for ligaments than it is for bone[2].

The elastic modulus or Young's modulus is the amount of stress needed to produce a given strain (the ratio of stress/strain) which is an inherent property of any material. It is also known as the stiffness of a material. From a biomechanical perspective, the com-

Table 1 Six potential movements of spine regarding three-dimensional axes.

	X-axis	Y-axis	Z-axis
Rotation	Flexion-extension	Right-Left axial rotation	Right-Left lateral bending
Translation	Lateral translation	Compression-distraction	A-P translation

bination of the NZ and the EZ is the range of motion (ROM) of the spine. However, clinically, the ROM refers to the maximum range the subject can move through without pain. Once the maximum limit of EZ is reached, any further stress results in a permanent, or plastic deformity. This zone is termed the plastic zone (PZ). The maximum limit of the elastic zone is also known as "yield point". If damage occurs at the ligament or join capsules between FSUs, the spine segments reach a state of relative laxity, with an increased NZ. The increase in the NZ is also known as "segmental instability"[1].

Definition of stability and instability

Spinal stability has been defined through knowledge gained via biomechanical studies. White et al[3] defined spinal stability as, "the ability of the spine under physiologic loads to limit patterns of displacement in order not to damage or irritate the spinal cord and nerve roots, in addition, to prevent incapacitating deformity or pain caused by structural changes." Similarly, the American Academy of Orthopedic Surgeons defined stability as, "The capacity of the vertebrae to remain cohesive and to preserve normal displacements in all physiological body movement,"[4].

Clinical instability is not an all-or-none phenomenon, commonly occurring on a spectrum ranging from stable to overtly unstable[1]. In biomechanics, the instability can be abnormal in either quality (instrumental failure, or abnormal coupling patterns) or quantity (increased motion). There are several testing options to investigate the underlying mechanics of the spine. Each test offers different information, so the appropriate test is dependent on the experimental goals.

Biomechanical Testing Methods

1. Strength testing (Load-to-failure)

Strength testing is also known as load-to failure testing. This study allows for the assessment of how much force is required for spinal or instrumental failure. The test determines the load-bearing capacity of the construct. In load-to-failure testing, the applied load is gradually increased until the spine or spinal construct fails. Photography and motion capture systems, i.e. electronic equipment that can analyze stress and strain at the point of failure, may be employed to assess the characteristics of failure. From this data, the researcher can generate a load-deformation curve and measure biomechanical parameters such as stiffness or ultimate strength (the highest load endured during failure). In addition, the way in which failure occurs (e.g., screw pullout or screw breakage) is also an important qualitative observation and maybe discernible from the acquired data[5].

Strength testing is therefore useful to study the mechanisms of spinal failure in spinal trauma, as well as the integrity of spinal instrumentation.

Testing model: Different testing models can be used to answer different research questions. Two common models are cadaveric tissue models and synthetic models.

Example of research questions:

Using cadaveric models; The study investigated which part of the cervical spine was more susceptible to rotational failure. Thus, the researcher loaded the entire cervical spine to failure during torsion. The specimens always failed at C1-2 first. After the failed level was removed and the remaining cervical spine was re-tested, the remaining specimens failed at a higher load in the lower cervical region[6].

Using synthetic models; They are commonly used to determine the strength of a fixation device without having confounding factors from cadaveric tissue. For example, one study[7] used plastic pucks as model of vertebrae to compare 12 different types of pedicle screw-rod and pedicle screw-plate construct.

2. Endurance testing (Fatigue testing)

Spinal fixation implants can loosen or fail when they are subjected to repetitive forces, eventually leading to failure from fatigue. Endurance or Fatigue testing is used to determine the hardware susceptibility to damage from fatigue. In fatigue testing, the construct is usually cycled until it fails or until a clinically relevant number of cycles are applied. The average spine is exposed to one million to three million cycle per year[8]. Spinal implants are expected to maintain stability until fusion is achieved, which generally occurs about six months after surgery. In this time, substantial cyclical loading can occur. Therefore, implants are typically expected to withstand around one million cycles of a normal load. Varying loads can be applied to establish a load-fatigue relationship curve.

Fatigue tests should be taken into consideration when designing and testing implants, however, the ex vivo nature of cadaveric specimens (implant-bone interface) is predisposed to break down, as it lacks the remodeling capabilities of in vivo tissue. Therefore, fatigue testing using a cadaveric model can only assess early implant behavior and characteristics of failure.

Fatigue tests can be classified as either: 1) Intermediate fatigue testing, or 2) Fatigue-to-failure testing. The decision to use one test vs the other depends on study design.

Testing models:

1. Intermediate fatigue testing is used as a part of flexibility testing. Flexibility tests are performed before and after fatigue is induced to show how much stability is lost because of fatigue. This type of fatigue is usually tested in animal and human cadaveric models as a part of flexibility testing.

2. Fatigue-to-failure testing typically uses synthetic models (plastic pucks), which are also used in load-to failure testing. In fatigue testing, a load is selected that is below the ultimate strength of the construct, determined by load-to failure testing. The model may be cycled at the rate as high as 20 cycles per second, depending on the standardized protocol being used. Cycling continues until failure or until a pre-chosen maximum number of cycles is reached (usually 1-10 million cycles)

Example of research questions:

1. Intermediate fatigue testing: Crawford et al.[9] studied the differential biomechanical effects of injury and wiring at C1-2. The human cadaveric C0-6 spines were tested via flexibility testing immediately after injury, after posterior cable and graft fixation (C1-2 wiring alone), and after 6,000 cycles of fatigue testing. Study outcomes supported the authors' conclusion that an interspinous cable-graft construct (C1-2 wiring alone) should be accompanied by an adjunctive stabilizing technique such as transarticular screw or a halo brace to ensure C1-2 fusion.

2. Fatigue-to-failure testing: Cunningham et al[7] utilized fatigue-to-failure testing in tandem with load-to-failure testing, using a synthetic model, to compare 12 different types of pedicle screw-rod and pedicle screw-plate constructs. In fatigue-to-failure assessments, the different types of constructs were cycled for 1 million cycles at three different peak loads to determine which type of constructs had adequate

fatigue strength to endure each load.

3. Flexibility testing (Stability testing)

In flexibility testing, loads are applied to a cadaveric spine while corresponding displacements are recorded. Applied loads, which are designed to be well below maximum allowable in vivo loads, are applied in several directions of loading (e.g. flexion, extension, lateral bending, and axial rotation). Additionally, these directional loads can be applied across several surgical conditions in the same specimen. To acquire the range of motion data, each spinal segment is instrumented with optical markers. Three-dimensional motion measurements (e.g.Optotrak 3020 system, Northern Digital, Waterloo, Ontario, Canada) are recorded automatically at 2 Hz.(Fig. 3) Custom software then converts marker coordinates into individual local coordinate systems for each vertebra so that intervertebral angular motion can be calculated[10]. Surface strain can also be recorded by surface strain gauges or digital image correlation and tracking (DIC).

Testing models: Human cadaveric or animal models are typically used in spinal biomechanical testing. Human cadaveric specimens are favored because their anatomy and tissue properties are easily generalized, elucidating spinal biomechanical mechanisms that translate well to living human subjects.

Example of research questions: Many research questions are about how much the spine moves during different injuries or instrumentation conditions. For example, in one study the researchers aimed to determine the relative amounts of movement at C1-C2 after instrumentation with various combinations of one or two transarticular screws and a posterior cable-secured graft[11]. The human cadaveric Occiput-C3 specimens were loaded nondestructively with pure moments to induce flexion, extension, lateral bending, and axial rotation while the range of motion data was acquired from optical markers attached to specimens.

4. Modeling

The biomechanical tests described earlier are usually enough to demonstrate the mechanical response of the spine to varying directional loads. However, in some cases, questions about biomechanical

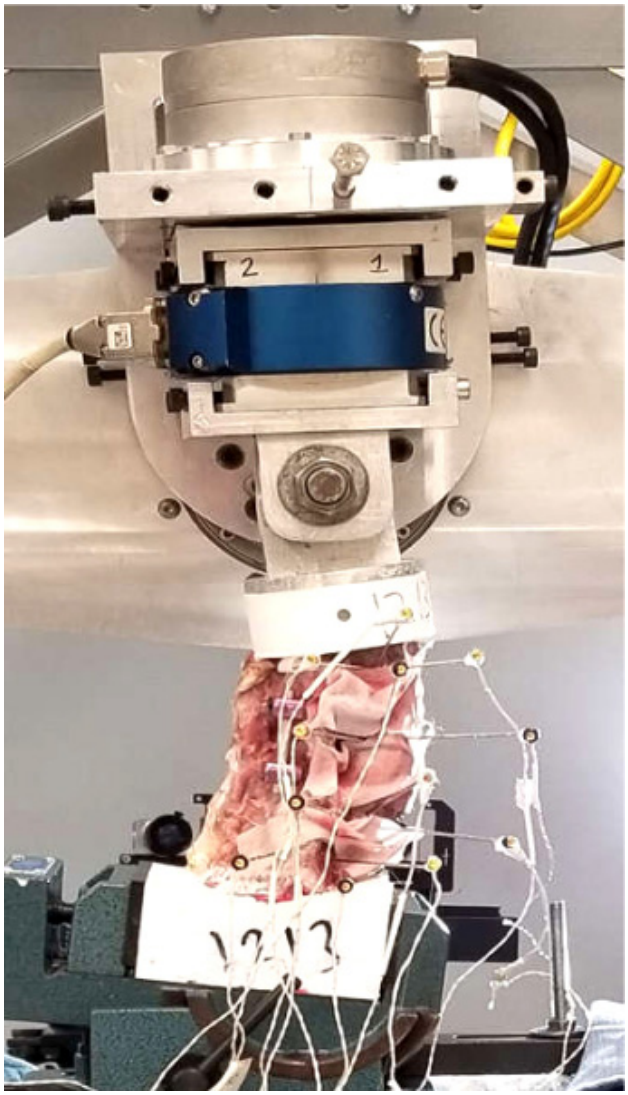

Fig. 3 Flexibility testing via the robotic testing frame of the instrumented lumbosacral spine with an optical marker attached. Three-dimensional motion measurements are recorded while testing. (Courtesy of Spine biomechanics lab, Barrow Neurological Institute, Phoenix, AZ, USA)

behavior may be impossible or too expensive to study using common tests. For example, a researcher may want to study a specific spinal deformity, osteoporotic model, or specific ligament effect while preserving other specific ligaments. In such cases, mathematical modeling or finite element analysis can be used as more feasible alternatives.

Finite-element modeling and analysis is the most common computational method for modeling spinal biomechanical behavior. Finite element modeling was originally intended for use in structural engineering. Now it is commonplace in many fields and has been used in spinal biomechanical research for over two decades. Finite element modeling spatially discretizes an input spinal geometry into many cells, or "elements" to generate a mesh. These elements interact with each other at junctions called "nodes"[12]. This technique can be used to simulate a variety of clinical situations by allowing the stresses, strains, and forces at any given location to be calculated using a computer.

Testing model: Computerized mathematical model.

Example of research questions: Zhao et al.[13] in-

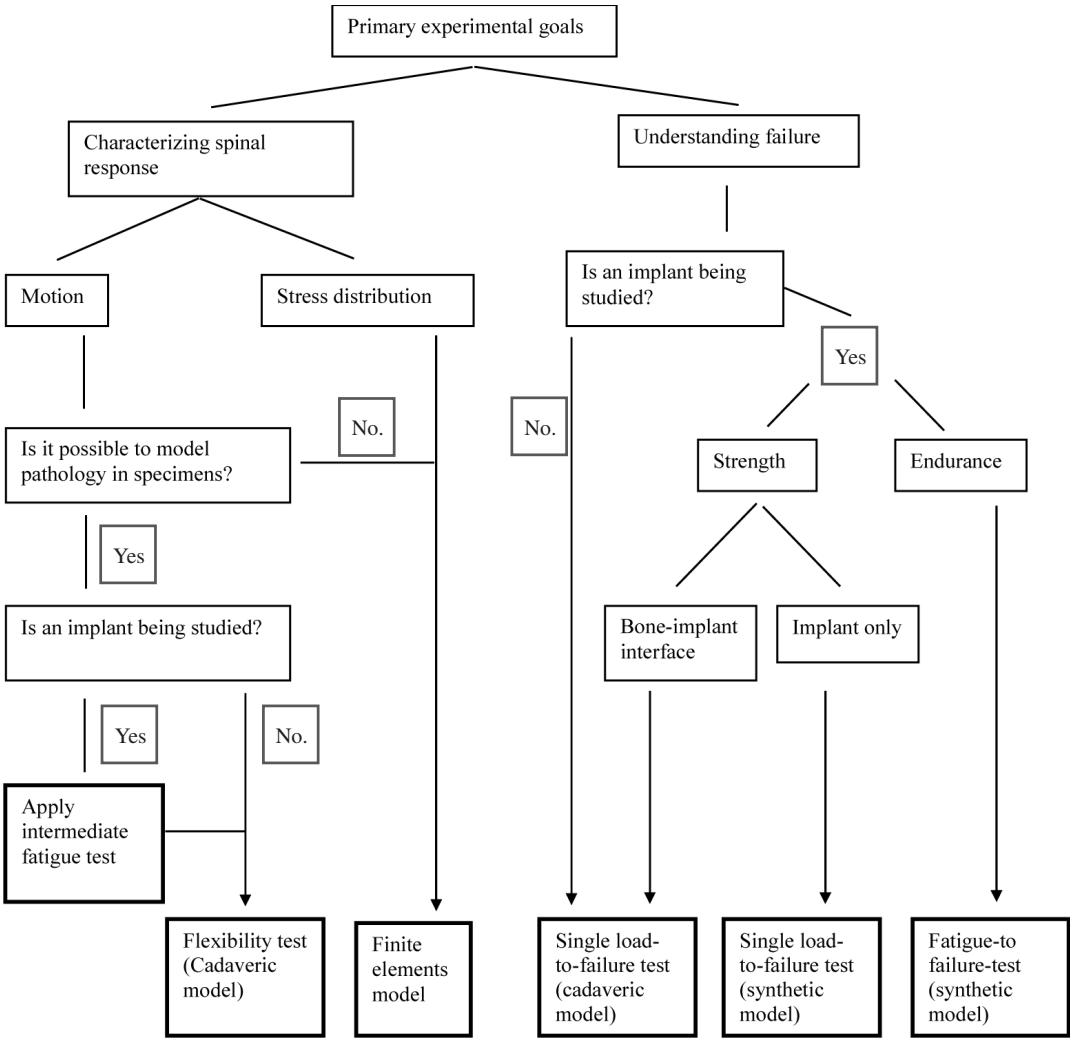

Fig. 4 Algorithm for choosing an biomechanical test appropriate to determine the desired information. (Adapted from Crawford NR. Advances in the understanding of spinal biomechanics through experimental research. Barrow Quarterly. 2002;18:4-10.)

vestigated the effect of different lordotic angles on the biomechanical behavior of the lumbar spine after TLIF in L4-L5 fusion. Finite element models were designed to test with different particular angles (57, 52, 47, and 40 degrees). The study result showed that the decrease in fused level lumbar lordosis generally increased in adjacent segment range of motion.

Conclusion

Spinal biomechanics is the fundamental knowledge required for understanding most areas of spinal surgery. A comprehensive understanding of biomechanics has also been used to develop novel advanced technology in spinal instrumentation. Thus, it is worthwhile for a spine surgeon to understand basic biomechanics and biomechanical tests. There are many options for testing in biomechanics including strength tests, flexibility tests, fatigue tests and finite element computational modeling. Each technique offers different information, therefore a researcher should select the appropriate technique to serve their experimental goals (Fig. 4).

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