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Review

Fundamental biomechanics of the spine—What we have learned in the past 25 years and future directions

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ABSTRACT

Since the publication of the 2nd edition of White and Panjabi's textbook, Clinical Biomechanics of the Spine in 1990, there has been considerable research on the biomechanics of the spine. The focus of this manuscript will be to review what we have learned in regards to the fundamentals of spine biomechanics. Topics addressed include the whole spine, the functional spinal unit, and the individual components of the spine (e.g. vertebra, intervertebral disc, spinal ligaments). In these broad categories, our understanding in 1990 is reviewed and the important knowledge or understanding gained through the subsequent 25 years of research is highlighted. Areas where our knowledge is lacking helps to identify promising topics for future research. In this manuscript, as in the White and Panjabi textbook, the emphasis is on experimental research using human material, either in vivo or in vitro. The insights gained from mathematical models and animal experimentation are included where other data are not available. This review is intended to celebrate the substantial gains that have been made in the field over these past 25 years and also to identify future research directions.

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1. Introduction

Clinical problems of the human spine continue to be prevalent in our society. Examples include low-back pain, sciatica, spinal deformity in both adults and children, spinal tumors, and spinal injury, including trauma to the spinal cord. Given that these clinical problems remain largely unsolved and that the spine plays an important mechanical role in human function, it is thus not a surprise that biomechanical research on the spine has expanded at a rapid pace. A PubMed search in June 2015 with the search terms 'spine' and 'biomechanics' showed that the number of articles in this field has increased exponentially over the past 25 years.

The classic textbook, Clinical Biomechanics of the Spine by White and Panjabi, was last published in 1990 and the next edition of this book is in the final stages of preparation. In the preparation of this third edition, we have had the opportunity to conduct a detailed literature review on the salient biomechanics literature related to the human spine over the past 25 years.

The purpose of this manuscript is to review what we have learned over the past 25 years in regards to the fundamentals of spine biomechanics. The material is organized in three main areas—the Whole Spine, the Functional Spinal Unit, and the Spinal Components (e.g. vertebra, intervertebral disc, spinal ligaments). My approach will be to briefly review what we knew in 1990, to outline what we have learned since that time, and to suggest areas for future research. Detailed reviews of papers are not provided, but classic references on a topic along with key new manuscripts are included for the reader to review. Due to space limitations, the spinal components of rib cage, muscle, spinal cord, and nerve root are not addressed in detail, nor is a review of in vivo spine kinematics or mechanobiology included. Further, topics such as the clinical biomechanical aspects of spinal trauma, spinal deformity, and surgical devices, techniques and instrumentation used in spine surgery are not included.

2. Whole Spine

The Whole Spine consists of the vertebrae of the cervical, thoracic, lumbar, sacral and coccygeal regions along with the intervertebral discs, ligaments, rib cage, and spinal musculature. In addressing the Whole Spine, a global view is taken towards spinal function rather than the local view when we address specific features of a Component such as the annulus fibrosus of the intervertebral disc. A summary of the topics addressed for the Whole Spine is shown in Table 1.

2.1. *Quantitative anatomy*

In 1990, we had a basic appreciation for the three-dimensional anatomy of the entire spine, with the obvious lordotic curves in the cervical and lumbar regions and the thoracic kyphosis. However, quantification of the global spine anatomy was not described in detail (Table 1).

Over the past 25 years, a detailed anatomic description of the entire spine was introduced that includes descriptions of the spine with respect to the pelvis and the hip joints. A new parameter, termed 'sagittal balance', has arisen. With respect to the pelvis, the notion of pelvic incidence and sacral slope (Fig. 1) provided a succinct way to quantitatively describe the position of the pelvis as it relates to the hip joints and to the sacrum (Duval-Beaupère et al., 1992; Legaye et al., 1998; Jackson et al., 1998; Roussouly et al., 2011). This pelvic geometry is also important in positioning the whole spine since the pelvis serves as the foundation upon which

Table 1
Whole Spine summary. This table presents an overview of our past (up to 1990) and current (1990–2015) knowledge on the key topics related to the Whole Spine. Possible future research directions are noted in the right-hand column.

	Up to 1990	1990–2015	Looking towards the future
Whole Spine — anatomy	i. Basic anatomy of the whole spine (i.e. cervical and lumbar lordosis and thoracic kyphosis) ii. Little quantitative anatomical data	i. Anatomical relationship between the spine, pelvis, and hip joints and the 'Sagittal Balance' concept ii. Considerable quantitative anatomical data	i. Explore spino-pelvic relationships in terms of the fundamental biomechanical principles (motion, stability and mechanical loads) ii. Develop normative anatomic databases for the spino-pelvic anatomy, including diurnal variations iii. Consider other possible anatomic principles that explain spinal function
Whole Spine — system	i. The three basic goals of the whole spine (load bearing, motion, neural protection)	i. Spine Stabilizing System hypothesis	i. Spine Stabilizing System hypothesis needs to be addressed clinically ii. Alternative principles upon which the spine functions should be considered
Whole Spine— loading	i. Compressive forces in the lumbar spine ii. Cervical intradiscal pressures	i. Predictions of in vivo lumbar compressive and shear forces and bending moments from math models ii. Thoracic intradiscal pressures; no new cervical load data	i. Further description of lumbar shear, bending, and torsion loads ii. Additional data on thoracic and cervical loads iii. Loads at the junctional regions of the spine (e.g. cervico-thoracic, thoraco-lumbar)
Mechanical stability	i. Low mechanical stability of the spine	i. Key role of spinal musculature in maintaining spine stability	i. What clinical problems can be addressed with advanced math. models?

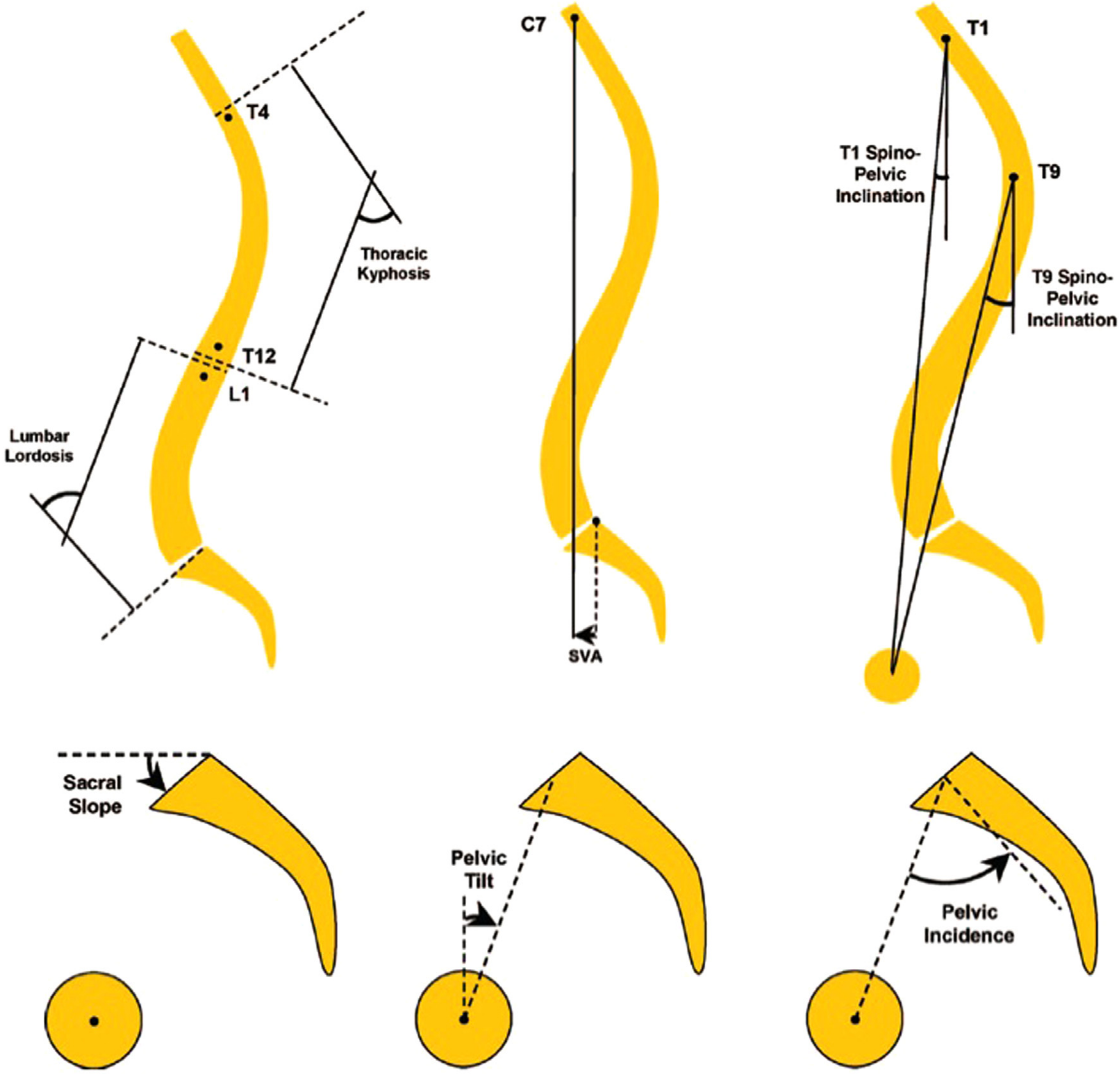

Fig. 1. Images from the article by Lafage et al. (2009) showing parameters used to describe the geometry of the spinal column (top three images) and the pelvis (bottom three images). The sagittal vertical axis (SVA) is what is commonly referred to as 'Sagittal Balance'. (Reprinted from Lafage et al., 2009, with permission from Wolters Kleuver.)

the spine stands. The sagittal balance is the anterior–posterior position of C7 with respect to the sacrum (Fig. 1) and this parameter has been shown to correlate with clinical symptoms (Glassman et al., 2005; Lafage et al., 2009).

2.2. Functional biomechanics

The biomechanical goals of the whole spine system—to provide structural support, to enable trunk movement, and to protect the neural elements—were well known in 1990. Further, the basic mechanisms by which the spine supports load and enables movement (i.e. for muscles to balance all external loads to the spine) were described at that time (Chaffin, 1969; Schultz and Andersson, 1981). The intervening 25 years has not yielded a single unifying principle upon which the spine has been shown to function. However, it has become clear that the body does attempt to stabilize the spine in addition to satisfying equilibrium (Crisco and Panjabi, 1992a; Cholewicki and McGill, 1996; Hodges and Richardson 1996; Radebold et al., 2001; van Dieën et al., 2003). The Spine Stabilizing System hypothesis (Panjabi 1992a, 1992b) was an attempt to connect the passive characteristics of the osteoligamentous spine with the active neuromuscular system. The basic premise of the hypothesis is that the human spine needs to be kept mechanically stable at all times to avoid injury that eventually leads to pain and that the maintenance of this mechanical stability is the role of the complex neuromuscular system. While this concept remains hypothetical, it is clearly a fruitful area for further research (Table 1).

The compressive loads in the lumbar spine were generally known in 1990, based upon the classic intradiscal pressure measurements of Nachemson, Andersson and colleagues (Nachemson, 1960, 1966; Andersson and Örtengren, 1974; Andersson et al., 1977). Since then, the lumbar intradiscal pressures have been replicated independently by Wilke et al. (1999, 2001) and Sato et al. (1999). Overall, these data generally support the loading patterns described by Nachemson and Andersson. One group has measured intradiscal pressures in the thoracic spine under a range of postures, observing similar patterns to those of Nachemson (Polga et al., 2004). Of note is that there remains only one study for cervical spine intradiscal pressures (Hattori et al., 1981). Therefore, additional research on cervical and thoracic spine loading would be excellent additions to the current literature (Table 1).

Our knowledge of in vivo spine loading is mainly for axial compressive forces. Shear forces across an intervertebral disc cannot be estimated from an intradiscal pressure since shear causes very little pressure change within the disc (Frei et al., 2002).

Shear loading in the spine and forces in the spinal musculature are thus predicted mainly from mathematical models (McGill, 1992; Marras and Granata, 1997; El-Rich et al., 2004; Shirazi-Adl et al., 2005; Kingma et al., 2007; Arjmand et al., 2009). However, it is important to keep in mind that validation of these predictions remains an ongoing challenge.

One major advance in the past 25 years in regards to in vivo spine loading has been the development and application of telemetry-based instrumented spinal fixators by Rohlmann and colleagues at the Charité Hospital in Berlin. By applying strain gauges to various implant components, they are able to record three-dimensional forces and moments on spinal implants in human patients across a range of activities. They have used this technology for anterior and posterior spinal devices (Rohlmann et al., 1994, 1997, 2007, 2008). Most impressive is that the group has made their data freely available on their website (www.orthoload.com) such that interested readers can explore any patient under any activity.

These instrumented implants have been used for various joints in the body including the hip, knee, and shoulder (Bergmann et al., 1993). In these instances, the implant loads can be used to determine precisely the forces and moments to which the respective joint, surgically treated with a joint replacement, is subjected. Unfortunately, this is not as simple for the spine. For both anterior and posterior spinal fixators, the implant is sharing load with the remaining components of the spine at that level. In the case of an anterior corpectomy device (Rohlmann et al., 2008), the implant will share load with any posterior structures, including the facet joints and/or any spinal instrumentation. For a posterior pedicle-screw implant (Rohlmann et al., 1997), the device will load share with the anterior column, most notably the intervertebral disc. Therefore, it can be challenging to use these data to infer the overall in vivo loading at any intervertebral level. Mathematical modeling of subjects with these telemetrized fixators may be an excellent opportunity to connect the implant loads with the loading across the entire intervertebral level.

The term 'stability' of the spine has been often misused and in the past, there remained a lack of clarity. The main challenge appears to have been the desire to apply a rigorous engineering term to clinical situations, where subjective parameters such as pain and function are the most important outcomes. The clear definition of'clinical stability' from White and Panjabi that dates back to 1978 was an important benchmark, but it remains difficult to apply clinically.

Substantial research over the past 25 years has shed light on the notion of mechanical stability in the spine such that in 2015, we should be able to be very specific with our terminology. Mechanical Stability is defined as the ability of a structure to return to its original state after being subjected to a perturbation. A mechanically unstable structure will experience local buckling under a compressive load, and it is likely that this is germane to clinical problems of the human spine (Reeves et al., 2007). The early research of Lucas and Bresler (1961) and Bergmark (1989) highlighted that the human spine was not mechanically stable without the active involvement of the spinal musculature. We now know that the human spine is mechanically unstable at very low compressive loads—less than 80 N for the lumbar spine (Crisco et al., 1992a, 1992b) and less than 12 N for the cervical spine (Panjabi et al., 1998). The critical role of the neuromuscular system was highlighted in the Spine Stabilizing System hypothesis (Panjabi, 1992a). The concept of mechanical stability has been subsequently incorporated into mathematical models that describe the basic functioning of the spine system (Kiefer et al., 1997; Kavcic et al., 2004; Shirazi-Adl et al., 2005; El-Rich et al., 2004; Arjmand et al., 2009; Stokes et al., 2011). Further, the concept has been used in defining and assessing rehabilitation exercises for people with low back pain (Vezina and Hubley-Kozey, 2000; McGill, 2001; McGill and Karpowicz, 2009). While all of the details surrounding the role of mechanical stability as it relates to clinical problems of the spine have not been worked out, this remains an active and fruitful area of research (Table 1).

Mathematical models of the spine have been developed using a wide range of approaches. These include simple equilibrium models, EMG-assisted models, optimization-based models, musculoskeletal dynamic models, and finite element models of various complexity (Cholewicki et al., 1995; Cholewicki and McGill, 1996; McGill and Norman, 1986; Marras and Granata, 1997; Keller et al., 2005; Shirazi-Adl et al., 2005; Arjmand et al., 2009; Han et al., 2012; Christophy et al., 2012). These models have enhanced our understanding of the spine and will continue to do so in the future, especially with the increasing power of computers. However, an ongoing issue with any mathematical model relates to its validation, which requires a comparison of model predictions with the in vitro and in vivo measurements that are made from other representative experiments. The focus of this manuscript is on our understanding of the human spine, based upon experimental data.

2.3. *Whole Spine—future steps*

There are many exciting avenues for future research with respect to the entire spine, some of which are highlighted in Table 1. Is there an overriding biomechanical principle that governs spine function, such as the Spine Stabilizing System hypothesis, and can that help explain clinical problems? Can we get even more accurate estimates of spine loading, including shear, torsion, and bending loads at all levels of the spine? Can we describe more precisely the unique characteristics of the junctional regions—e.g. cervicothoracic, thoracolumbar, lumbosacral—and link this behavior with clinical problems? Can validated mathematical models of clinical conditions predict the outcome of various therapies? This would surely be an exciting development!

3. Functional spinal unit

The functional spinal unit (FSU) is the basic building block of the spine, consisting of two adjacent vertebrae, the intervertebral disc, the facet joints, and the spinal ligaments. In this section, I address the physical properties and functional biomechanics of the FSU through experimental studies on human samples and I include relevant data from studies on multi-FSUs. The topics presented with respect to the FSU are summarized in Table 2.

3.1. *Physical properties*

The basic physical properties of the human FSU in 1990 were based upon the classical studies for the cervical (Moroney et al., 1988), thoracic (Panjabi et al., 1976), and lumbar (Farfan, 1973; Liu et al., 1975; Berkson et al., 1979; Tencer et al., 1982; Posner et al., 1982; McGlashen et al., 1987) regions of the spine. The concept of main and coupled motion behavior was described, with the major emphasis on axial rotation–lateral bending coupling in the cervical and lumbar regions (Panjabi et al., 1976; Pearcy and Tibrewal, 1984; Panjabi et al., 1989a). The compressive FSU behavior was well-characterized from the classic work of Nachemson, Hirsch, and others (Hirsch and Nachemson, 1954; Hirsch, 1955). A three-dimensional stiffness matrix for a thoracic FSU was described in 1976 (Panjabi et al., 1976). Most of these studies reported the FSU properties as linear, even though the first description of the neutral zone parameter as a way of helping describe the non-linear behavior of the FSU was already reported (Panjabi et al., 1982).

Table 2
Functional Spinal Unit Summary. This table presents an overview of our past (up to 1990) and current (1990–2015) knowledge on the key topics related to the functional spinal unit. Possible future research directions are noted in the right-hand column. Short forms used: FSU for functional spinal unit, BMD for bone mineral density, DD for disc degeneration.

	Up to 1990	1990–2015	Looking towards the future
FSU physical properties	i. Main and coupled motions for cervical, thoracic and lumbar FSUs ii. FSU load–displacement curves as linear iii. Effects of compressive preload/ muscle action iv. Basic stiffness matrix for the thoracic FSU	i. Further detail on main and coupled FSU behavior ii. Non-linear FSU load–displacement curve, with Neutral Zone parameter iii. Stiffening effect of compressive preload on the FSU; Follower Load concept iv. Lumbar FSU stiffness matrix	i. Cervical and thoracic shear characteristics ii. Can non-linear features of the FSU be detected in vivo? iii. Stiffness matrices that reflect the non-linear behavior of FSUs iv. More data on FSU dynamic behavior
FSU—loading	i. Basic mechanisms of load transfer across a FSU in compression, shear, bending, and torsion ii. Common FSU injury mechanisms for end-plate fracture and disc herniation	i. Stress profilometry technique provided detail on load transfer across the FSU in compression ii. Further studies on FSU failure mechanisms	i. Compressive load-sharing in the cervical FSU ii. Details regarding mechanisms of disc herniation
FSU—effect of aging	i. FSU response with increasing DD not well understood ii. FSU strength in compression known to decrease with advancing age and lower BMD	i. Effect of DD on FSU load–displacement behavior ii. Inter-relationship between BMD and DD on FSU responses in load bearing and motion	i. Effects of aging in terms of DD and BMD should be accounted for in future FSU experiments and math modeling studies ii. Connect in vitro FSU properties with in vivo observations
FSU—effect of injury	i. Effects of acute disc, ligament, and facet injury on FSU behavior	i. Additional studies shed further light on the effects of various injuries.	i. Cellular responses to injury

In the past 25 years, many studies have expanded our knowledge of the physical properties of the FSU, with a focus on directions other than axial compression. We now know that the bending behavior of the FSU is non-linear in flexion–extension (Fig. 2b) and lateral bending and almost linear in lumbar axial rotation. These observations have been made in studies on the cervical (Fig. 2a) (Wen et al., 1993; Panjabi et al., 2001b; Nightingale et al., 2002, 2007) and thoracolumbar (Oxland et al., 1992; Oda et al., 2002; Stemper et al., 2010; Panjabi et al., 1994; Wilke et al., 1994; Heuer et al., 2007) regions. The shear behavior of the lumbar spine is roughly linear (Fig. 2c) (Frei et al., 2001, 2002; Gardner-Morse and Stokes, 2003; Lu et al., 2005; Skrzypiec et al., 2012; Schmidt et al., 2013), with some investigators describing the anterio-posterior shear behavior as bi-linear due to facet contact (Gardner-Morse, 2004). The characteristics of the cervical and thoracic FSUs under shear loading is not well-described (Table 2).

An enhanced stiffness matrix was proposed for the lumbar spine by Gardner-Morse and Stokes (2004), which is important for some investigators attempting to mathematically model the spine. The challenges associated with these stiffness matrices in accounting for finite and non-linear structural behavior are considerable and they have been highlighted (O'Reilly et al., 2009).

The non-linear behavior of the FSU in bending led to the definition of the Neutral Zone (NZ) in addition to the Range of Motion (ROM). The NZ represents the low stiffness region of the load–displacement curve and is often referred to as the joint laxity, in contrast to the well-known ROM (Fig. 2a and b), which represents the total extent of motion under a given load (Panjabi et al., 1982, 1988, 1994). While the NZ has been quantified under a range of loads, an important characteristic of the NZ is its high sensitivity to injury, in comparison to the ROM parameter (Oxland and Panjabi, 1992; Ching et al., 1995; Zhu et al., 1999).

The biomechanical characteristics of the FSU under a superimposed compression force have been of interest due to the known muscle activity in the spine (Panjabi et al., 1977, 1989b). An important observation has been that bending motion decreases under a superimposed compressive force (Janevic et al., 1991; Wilke et al., 1995; Stokes and Gardner-Morse, 1995; Gardner-Morse and Stokes, 2003). This may be due to artifact loads being applied to the FSU (Cripton et al., 2000), to increased contact of the facet joints with added compression (Pollintine et al., 2004), and/or to increased annular stiffness under compression (Iatridis et al., 1998). A unique type of superimposed compressive load, termed the 'Follower Load', was described by Patwardhan and colleagues for the testing of multi-FSU specimens. By passing the compressive preload through the sagittal center of rotation at each vertebral level, limited stiffening behavior secondary to the compressive preload was observed (Patwardhan et al., 1999, 2003). As a result, this Follower Load is an effective way in which to simulate physiological compression in a laboratory environment (Fig. 3), since in a neutral posture the resultant of all muscle forces likely passes along a similar contour. However, it is worth noting that the Follower Load does not replicate the physiological action of the spinal musculature across a range of postures and movements.

The major role of the nucleus pulposus and annulus fibrosus in resisting compression (Hirsch and Nachemson, 1954; Hirsch, 1955; Nachemson, 1966) and the load-bearing role of the annulus in compression for intervertebral discs with advancing degeneration were noted in early studies (Perey, 1957). The load bearing ratio through the posterior elements was shown by Adams and Hutton (1980) to be posture dependent, being about 16% in a neutral posture and dropping to zero in a flexed posture. After 1990, our understanding of disc mechanics advanced with the development of a technique termed stress profilometry, in which a pressure transducer is pulled through the cross-section of a loaded intervertebral disc. This technique showed us very clearly how compressive load was distributed across the intervetebral disc of a FSU (McNally and Adams, 1992). The early studies in human cadaveric FSUs showed us the hydrostatic nature of the disc when it is healthy and how the central loading drops substantially when it becomes degenerated (Fig. 4). Further, in this degenerative state, the loading on the annulus rises dramatically (Fig. 4), which is consistent with the observations of Perey noted above (1957).

3.2. *Functional biomechanics*

While this review does not address spinal trauma as a general topic, it is worth addressing two common failure mechanisms of the FSU—namely endplate fractures and intervertebral disc

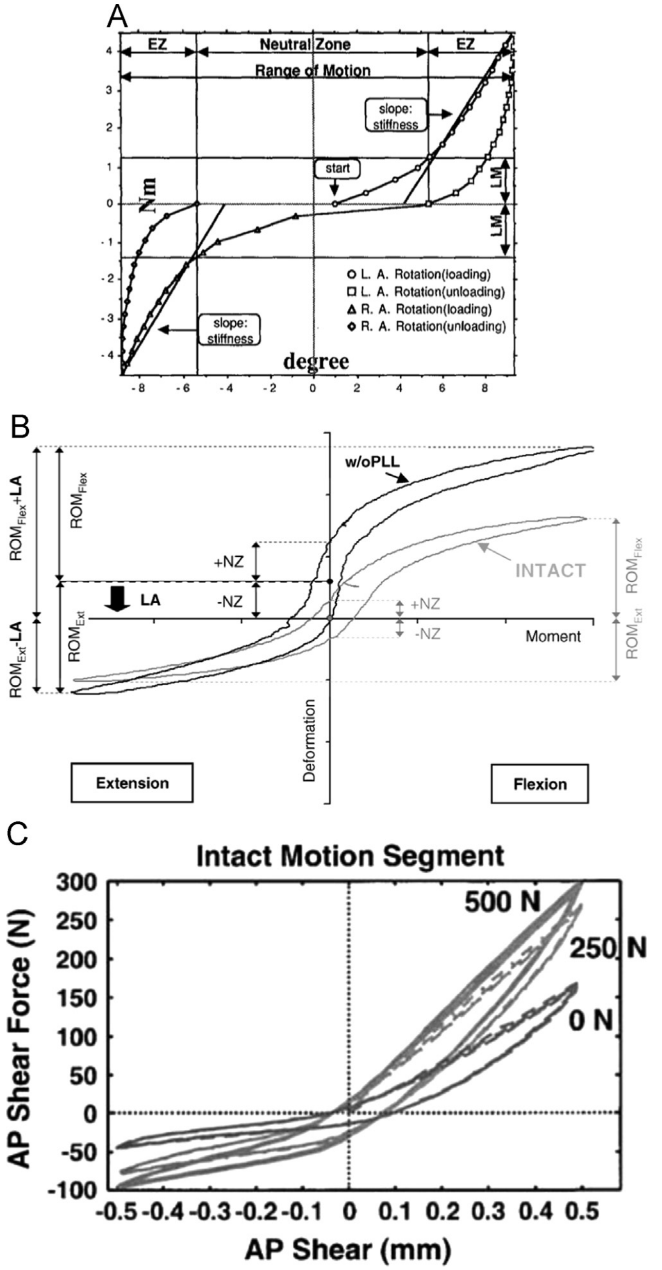

Fig. 2. Typical load–displacement curves for functional spinal units. (A) Moment–rotation curve for the cervical spine in axial rotation, depicting strong non-linear behavior. [Reprinted from Wen et al., 1993, with permission from Springer.] (B) Rotation–moment curve for the lumbar spine in flexion–extension, depicting non-linear behavior. [Reprinted from Heuer et al., 2007 with permission from Elsevier.] (C) Force–displacement curves for the lumbar spine in anterior–posterior shear under different compressive preloads. [Reprinted from Gardner-Morse and Stokes, 2004 with permission from Elsevier.].

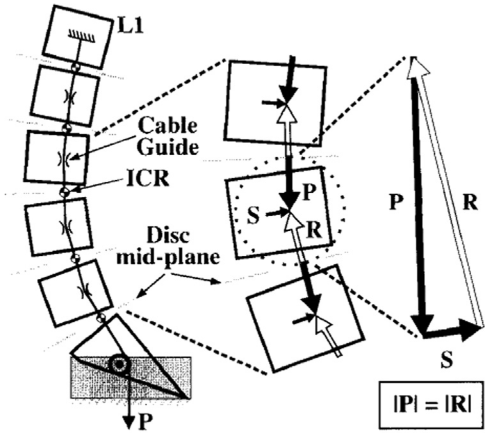

Fig. 3. Schematic diagram of the Follower Load shown on a lumbar spine at left with resolved forces on the right. [Reprinted from Patwardhan et al., 2003 with permission from Wiley.]

herniations. Both lesions are common clinically and thus an understanding of their etiology is relevant.

Under compressive loading of a FSU, the initial failure typically occurs at the vertebral endplate due to the high pressure in the nucleus pulposus in contact with the endplate (Perey, 1957; Brinckmann et al.; 1989; Yoganandan et al., 1988; Fields et al., 2010; Wade et al., 2014). The most common endplate to fail is the superior vertebral endplate (i.e. inferior to the disc) since it has been shown to be thinner and weaker than the inferior vertebral endplate (Zehra et al., 2015; Grant et al., 2001; Roberts et al., 1997). The creation of disc herniation in a laboratory setting was described first by Adams and Hutton in their classic 1982 paper that showed the importance of rapid, out of plane loading to produce disc herniations. Other investigators since that time have produced herniations with long-term cyclic loading or high-rate single impact (Adams and Hutton, 1985; Gordon et al., 1991; Callaghan and McGill, 2001; Veres et al., 2009; Wade et al., 2014). These studies and parallel mathematical models (Schmidt et al., 2007) demonstrated that a disc herniation is produced in situations where the annulus fibrosus is stretched first, often to the limit of physiologic motion, and then the disc is subjected to high compressive loads.

The biomechanical effects on the FSU of acute injury to the intervertebral disc were known at a nascent level in 1990. Early studies highlighted the relative effects of acute annulotomy and nucleotomy in the lumbar spine (Panjabi et al., 1984; Goel et al., 1985, 1986; Brinckmann and Horst, 1985; Brinckmann, 1986). In the past 25 years, a range of studies have documented several important features of acute disc injury. First, all injuries to the disc produce some biomechanical effects that include altered structural response and/or redistributed load between the components of the FSU (Brinckmann and Grootenboer, 1991; Frei et al., 2001; Meakin et al., 2001; Johannessen et al., 2006; Vresilovic et al., 2006; Przybyla et al., 2006; O'Connell et al., 2011a). As one would predict, the degree of these biomechanical changes is generally in proportion to the severity of the lesion. For example, a limited nucleotomy results in more subtle kinematic changes than a radical nucleotomy (Johannessen et al., 2006). Even the smallest lesions, however, may have a significant biological effect. As demonstrated first by the group in Adelaide, an annular stab incision results in degenerative changes in the disc of an animal (Osti et al., 1990). Subsequent studies on this topic demonstrated the consistency of this finding across different species and for even very small lesions such as needle puncture (Kaigle et al., 1997; Masuda et al., 2005; Michalek et al., 2010; Martin et al., 2013). Finally, animal studies on the healing of disc lesions have documented that natural annular repair does not substantially improve the impaired biomechanical response of the FSU (Ethier et al., 1994; Ahlgren et al., 1994, 2000), thus clinicians should attempt to minimize iatrogenic annular damage.

The effect of damage to the articular facets on biomechanical behavior of the FSU was not established until the decade of the 1990's. Beginning with the lumbar spine, Abumi and colleagues found that substantial motion increases were observed only when one facet was completely removed (Abumi et al., 1990), and this was supported in a computational model (Natarajan et al., 1999). Similar studies in the cervical spine found that removal of only

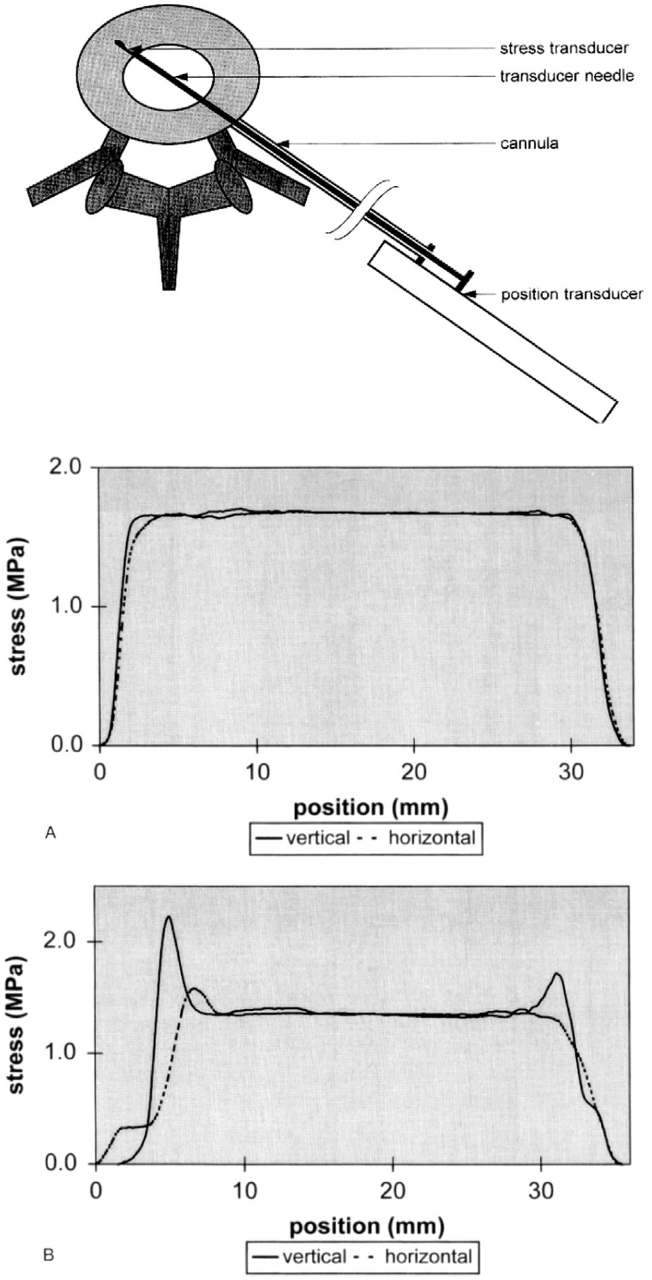

Fig. 4. Schematic diagrams of the Stress Profilometry approach, showing the experimental procedure (top image) and typical data for healthy and degenerated discs (bottom images) [Reprinted from McNally DS1, Shackleford IM, Goodship AE, Mulholland RC. In vivo stress measurement can predict pain on discography. Spine (Phila Pa 1976). 1996;21(22), 2580–2587, with permission from Wolters Kleuver].

25% of the medial facets can result in large motion increases (Zdeblick et al., 1993; Nowinski et al., 1993). These basic studies on the biomechanics of the facet joints provide tangible guidance for surgical strategies in the spine.

A wide range of studies have documented the effect of ligament injury on increases in spinal motion in the cervical (Panjabi et al., 1975), thoracic (White and Hirsch, 1971; Panjabi et al., 1981), and lumbar regions (Posner et al., 1982; Zander et al., 2004; Heuer et al., 2007). Many of these are classic studies that have influenced the surgical approach to spinal problems, especially spinal trauma.

Given that the loading distribution across an FSU under axial compression is altered substantially with disc degeneration, it is not surprising that several groups have described decreased compressive stiffness of the FSU with degenerated discs (Hirsch and Nachemson, 1954; Hansson et al., 1987; Yoganandan et al., 1988). A long-standing clinical question, first hypothesized by Knutsson in 1944 and further popularized by Kirkaldy-Willis and Farfan in 1982, has been whether disc degeneration that is manifest as LBP results in kinematic changes that may be observable by measuring relative vertebral motions. An early in vitro investigation by Nachemson and colleagues found no clear effects of disc degeneration on rotational motion (Nachemson et al., 1979). Given that clinical studies found clear evidence of less spinal motion with aging and that people with LBP had less motion than asymptomatic individuls (Dvorak et al., 1992, 1995), there appeared to be a contradiction in the data. A range of in vitro studies over the past 25 years focussed on whether the rotational motion of human FSUs in the lumbar spine differed with advancing disc degeneration. These studies determined some relationships between bending behavior and disc degeneration, but the results were rather subtle. In general, lumbar spine ROM decreases slightly with advancing degeneration in flexion–extension and lateral bending while it increases slightly in axial rotation (Mimura et al., 1994; Oxland et al., 1996; Fujiwara et al., 2000; Krismer et al., 2000; Kettler et al., 2011). Thus, in vivo observations of decreased motion in LBP subjects are likely due to a pain response and that the motion changes with aging are likely due to muscular changes rather than an inherent decrease in spinal column flexibility (Dvorak et al., 1992, 1995).

3.3. *Functional spinal unit—future steps*

The biomechanical characteristics of the FSU are fundamental for understanding the spine and its various pathologies. We have gained much knowledge on the structural response of the various regions of the spine, but there are some notable exceptions. There is a lack of data on the shear behavior of the cervical and thoracic FSUs, which is surprising considering that artificial replacement of the cervical joints is being done clinically. The non-linear characteristics of the FSU, notably the NZ, are sensitive to injury but in vivo measurement of these parameters is challenging. A notable intraoperative study demonstrated the NZ parameter in vivo (Hasegawa et al., 2008) and further studies of this kind will be important. A more accurate description of three-dimensional stiffness matrices is clearly needed for all regions of the spine, as some future mathematical models will need such information. We now understand that intrinsic variables such as bone density and disc degeneration affect profoundly the biomechanical behavior of an FSU and thus these parameters should be incorporated into experiments assessing physical properties and functional biomechanics of the spine. Further, mathematical models should incorporate these parameters such that the effects can be better understood. Finally, virtually all of the parameters addressed in this section on the FSU are quasi-static. The dynamic properties of the FSU (Crisco et al., 2007; Reeves and Cholewicki, 2010) should be a focus of future research efforts as they are likely very relevant to the in vivo situation (Table 2). Finally, the in vitro FSU motion patterns should be compared with the in vivo kinematic patterns that have been reported in recent years to enhance our understanding of spinal column–muscle interactions.

4. Component—vertebra

The vertebra consists of the vertebral body anteriorly, the neural arch posteriorly and a series of processes that serve as connection points for ligaments and muscles. The outer shell of the vertebra is mainly cortical bone and the inner region a network of cancellous bone. While we knew much in 1990 about the form and function of the spinal vertebrae, substantial research over the past 25 years has shed additional light on the biomechanics of the vertebra. A summary of the topics addressed for the Vertebra is shown in Table 3.

Table 3
Vertebra summary. This table presents an overview of our past (up to 1990) and current (1990–2015) knowledge on the key topics related to the Vertebra. Possible future research directions are noted in the right-hand column. Short forms used: BMD for bone mineral density, CSA for cross-sectional area, DD for disc degeneration.

	Up to 1990	1990–2015	Looking towards the future
Vertebral anatomy	i. Basic anatomy of the vertebrae ii. Some quantification of vertebral anatomy available, e.g. facet orientation	i. Quantitative vertebral anatomy from all spinal regions ii. Average BMD data across the lifespan exist from population studies iii. Heterogeneous architecture of the vertebra	i. Detailed characterization of vertebral architecture needed, particularly cancellous bone throughout the lifespan
Vertebral strength	i. Vertebral strength increases from cranial to caudal ii. Effect of BMD on vertebral strength established iii. Relative roles of cortical shell and cancellous core to vertebral strength was in question	i. Vertebral body strength determined by BMD and CSA, with smaller role of internal architecture ii. Endplate strength varies substantially across its surface iii. Cortical shell and cancellous core contribute to vertebral strength iv. Loading rate effect on vertebral strength	i. Effects of non-axial loads on vertebral strength (e.g. flexion–compression) ii. Influence of DD on vertebral strength iii. Effect of dynamic loading on the vertebra important
Vertebra—effect of aging	i. Decreasing BMD with advancing age well-established	i. Effect of DD on vertebral endplate structure ii. Neural arch loading in compression increased with advancing age and DD, possibly leading to lower BMD	i. Interplay between DD and bony changes important ii. Further investigate the Neural Arch loading hypothesis and explore other possibilities

4.1. *Quantitative anatomy*

The anatomy of the vertebrae was in the early stages of being quantified in 1990. The external dimensions of the vertebra were reported by few authors (Nissan and Gilad, 1986; Berry et al., 1987; Scoles et al., 1988), but we now have a much more comprehensive description of the quantitative vertebral anatomy for the cervical, thoracic, and lumbar regions of the spine (Panjabi et al., 1991a, 1991b, 1992c, 1993; Frobin et al., 1997). The ethnic variation of these vertebral dimensions has been reported by some investigators (Tan et al., 2004) and this should continue. The cortical bone and endplate thicknesses in the vertebrae have been quantified (Edwards et al., 2001; Panjabi et al., 2001a). Further, we now know that vertebral bone is also heterogeneous within a single vertebra and a single spinal column (Grote et al., 1995; Ritzel et al., 1997; Banse et al., 2001; Thomsen et al., 2002; Silva et al., 1994; Fazzalari et al., 2006; Hulme et al., 2007).

The internal morphology and morphometry of the vertebrae has been described in great detail over the past 25 years, due largely to advances in medical imaging techniques such as CT and micro-CT. These methods that can quantify the bone mineral density (BMD) in a vertebra have demonstrated that the vertebral bone is highly variable across the population, leading to clinical definitions for osteoporosis and osteopenia (Riggs et al., 2008; Edwards et al., 2015).

4.2. *Physical properties*

The strength of the vertebral body in axial compression was shown to increase along the entire spinal column from cervical to lumbar by Messerer in 1880, due largely to increasing cross-sectional area. The critically important role of the vertebral bone density in the compressive strength of the vertebra was well known in 1990 (Atkinson 1967; Bell et al., 1967; Carter and Hayes, 1976; Mosekilde and Mosekilde, 1986) and it was shown by various groups that a product of BMD and vertebral cross-sectional area correlated highly with compressive strength (Hansson and Roos, 1980; Mosekilde and Mosekilde, 1986; Brinckmann et al., 1989). Since that time, the non-invasive imaging of the vertebra has enabled investigators to predict the vertebral compressive strength from measurement of the internal BMD and the vertebral size such that the correlation coefficients between predicted and known strength is about 80% (Mosekilde et al., 1989; Ebbesen et al., 1999).

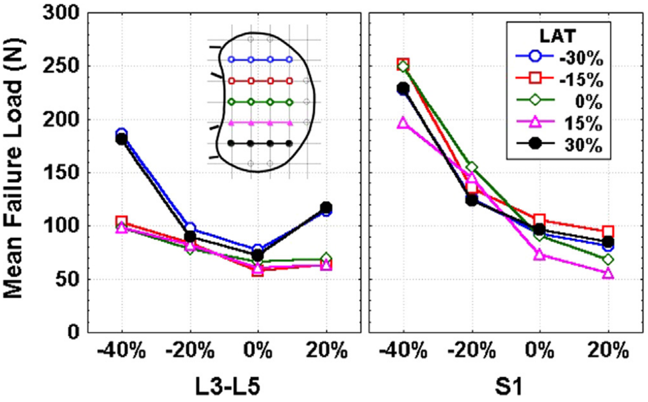

Fig. 5. Plot of the lumbar and sacral endplate strength profiles due to indentation testing, showing how the posterolateral regions are the strongest in the lumbar spine. [Reprinted from Grant et al., 2001, with permission from Wolters Kleuver.]

In 1990, there continued to be controversy regarding the relative load bearing roles of the cortical shell versus the cancellous core of the vertebral body (Rockoff et al., 1969; McBroom et al., 1985), with Rockoff suggesting an important load-sharing role for the vertebral cortex and McBroom et al. reporting a negligible role. Since that time, it has become clear that the vertebral cortex and the cancellous bone both contribute substantially to the strength of the vertebral body (Silva et al., 1997; Eswaran et al., 2006; Christiansen et al., 2011). It must be noted that some studies on this topic loaded the vertebra artificially by flat plates on their endplates, while actual vertebral loading is through an adjacent intervertebral disc. Such attention to physiological boundary conditions is clearly important for research in this area.

The bony endplates on the superior and inferior surfaces of the vertebral body were characterized by few investigators in 1990 (Whitehouse et al., 1971; Bernick and Cailliet, 1982). Over the past 25 years, the thickness and physical properties of these endplates has been described. Indentation studies on the lumbar vertebral endplates found that the posterolateral region of the endplate was about 2.5X stronger than the most central regions (Grant et al., 2001) with an average map of strengths shown in Fig. 5. These experimental results are consistent with observations in cervical and lumbar vertebrae that the endplate thickness is thickest in the posterolateral regions (Pitzen et al., 2004; Müller-Gerbl et al., 2008). With advancing disc degeneration, the strength of the central endplate decreases but it increases at the more peripheral sites (Grant et al., 2002), which is likely a response of the vertebral bone to an altered loading state where more load passes through the annulus fibrosus (Fig. 4).

While the vertebral strength in compression is primarily a function of bone density and cross-sectional area, other parameters do have an effect. An important mechanical parameter affecting vertebral strength is the loading rate, with well-known increases in strength at faster rates. High loading rates of 2.5 m/s doubled the vertebral strength measured at quasi-static rates (i.e. 10 mm/s), for example (Ochia et al., 2003). Variations in vertebral architecture other than a simple bone mineral density (e.g. trabecular spacing) do influence vertebral strength, but not to the degree of bone quantity (Lochmüller et al., 2008; Thomsen et al., 2013). Variables such as gender and vertebral level do not appear to have a significant effect on strength when the important size effect is taken into account (Singer et al., 1995).

4.3. *Functional biomechanics*

It is well-known that flexed postures result in more vertebral body loading and extended postures create more neural arch loading (Yang and King, 1984; Lorenz et al., 1983; Adams and Hutton, 1980). More recent research has highlighted that with advancing age, the percentage of load passing through the neural arch increases, particularly after about age 60 (Pollintine et al., 2004). This observation is largely felt to be caused by the decreasing height of the intervertebral disc with age, which eventually increases the contact forces in the neural arch (Adams et al., 2006).

Adams and colleagues suggest that with this increased neural arch loading with advancing age, there is a concomitant reduction in anterior column loading which, due to our knowledge of bone's response to mechanical loading, will result in a reduction in vertebral body bone density and is a possible explanation for vertebral compression fractures in the elderly (Adams et al., 2006). Future research should address this hypothesis in more detail, particularly exploring why more direct relationships between vertebral bone density and disc degeneration have not been observed previously (Shao et al., 2002).

4.4. *Vertebra—future steps*

There remains much exciting biomechanical research to be done in regards to the spinal vertebra. Better characterization of the bone in the vertebra is important, particularly as it relates to age-related changes and osteoporosis. Further exploration of the Adams hypothesis relating aging with altered loading patterns and bone changes is needed. Additional research on the interrelationship between the vertebra and the intervertebral disc will be important, given the role of the vertebra in disc nutrition (Table 3).

5. Component—intervertebral disc

The intervertebral disc consists of the central nucleus pulposus surrounded by the annulus fibrosus peripherally and the cartilaginous endplates rostrally and caudally. The structure and function of the disc is one of the most heavily researched biomechanical topics over many decades, beginning with the classic work of Hirsch, Brown and Virgin in the 1950s (Hirsch, 1955, Brown et al., 1957; Virgin, 1951). A summary of the topics addressed here for the Intervertebral Disc is shown in Table 4.

5.1. *Quantitative anatomy*

The basic anatomy of the intervertebral disc was known in 1990, including some quantitative data on disc heights along the spinal column (Pooni et al., 1986). Considerable research over the past 25 years, much of it rather recent, has highlighted the complex internal architecture of the disc. A vast network of connective tissue that joins the nucleus to the cartilaginous endplate, nucleus and the annular fibers, and interconnections between the annular fibers have been described recently for the ovine disc (Schollum et al., 2009; Wade et al., 2011, 2012). A major advance has been the imaging of the intervertebral disc using MRI, which enables visualization of its internal structure, including the nucleus pulposus (Weidenbaum et al., 1992; Marinelli et al., 2009).

The many qualitative changes that occur to both the nucleus and annulus with aging and degeneration were described before 1990 (Hirsch, 1956; Twomey and Taylor, 1987) and a validated subjective grading scheme for disc degeneration was described (Thompson et al., 1990). Since then, more detailed histologic

Table 4
Intervertebral disc (IVD) Summary. This table presents an overview of our past (up to 1990) and current (1990–2015) knowledge on the key topics related to the Intervertebral Disc. Possible future research directions are noted in the right-hand column. Short forms used: MRI for magnetic resonance imaging, IDP for intradiscal pressure.

	Up to 1990	1990–2015	Looking towards the future
IVD anatomy	i. Basic anatomy of the IVD understood ii. Little quantitative anatomical data available	i. Microstructural details of disc anatomy ii. Some quantification of disc anatomy, especially heights iii. MRI enabled visualization of disc internal structure	i. Further characterization of IVD anatomical details ii. Correlation of anatomic structure with MRI changes
IVD physical properties	i. Basic linear properties of disc in compression described ii. Creep behavior in compression iii. Physical properties of annulus fibrosus reported	i. More detailed disc physical properties ii. Viscoelastic behavior of disc described, including poroelastic behavior iii. More detailed annulus fibrosus and nucleus pulposus physical properties	i. Properties of cervical and thoracic discs not well characterized ii. Dynamic properties of the disc
IVD nutrition	i. Basic mechanism of diffusion that underlies IVD nutrition described	i. Additional studies on disc nutrition ii. Mathematical modeling of IVD nutrient diffusion	i. More detailed understanding of IVD nutrition, including through annulus fibrosus ii. The importance of the diurnal changes in IDP on disc nutrition are important to understand.
IVD damage	i. Basic IVD injury mechanisms of annular tearing and annular delamination described	i. Experimental and mathematical models enhance understanding of these injuries	i. Better understand role of annular tearing on Low back pain

changes in the disc with age have been documented to include a wide range of anatomic changes in the disc structure shown to begin in the second decade of life (Boos et al., 2002; Roberts et al., 2006). A grading scheme for disc degeneration from MRI scans was described by Pfirrmann et al. (2001, 2006).

An important development in the past 25 years was the description of the cervical intervertebral disc, where the authors noted that the structure of the annulus fibrosus was more crescent-shaped with thinning posteriorly, which is noticeably different from that of the thoracolumbar discs (Mercer and Bogduk, 1999).

5.2. *Physical properties*

The main biomechanical properties of the intervertebral disc were well established by 1990, including stiffness of the lumbar intervertebral disc in axial compression, shear, bending, and torsion. Differences in these stiffnesses from those of the FSU suggest the loading role of the facet joint in certain loading directions (e.g. anterior/posterior shear). Many additional studies on these properties have been conducted in the subsequent 25 years. Of note is that few studies exist that measured the tensile stiffness of an intervertebral disc, which is surprising given that distractive injuries to the spine do happen, albeit less frequently than in compression. Further, comparatively few studies have evaluated the physical properties of the cervical or thoracic discs (Table 4).

Some basic viscoelastic properties of the intervertebral disc were known before 1990. Kazarian documented the basic creep characteristics of lumbar intervertebral discs and showed that degenerated discs deformed more rapidly than normal discs (Kazarian, 1975). A three-element solid Kelvin model (i.e. spring in series with a spring-dashpot) was shown to be effective for modeling the viscoelastic behavior of the disc (Burns and Kaleps, 1980; Keller et al., 1987). Since 1990, a series of additional experiments have provided supportive data on the viscoelastic characteristics of the lumbar intervetrbal disc. For example, the stiffness in all six degrees of freedom was shown to be sensitive to loading frequency, with up to 83% stiffness increase across frequencies from 0.001 to 1.0 Hz (Costi et al., 2008). Studies on loading rate show large increases in disc stiffness across several orders of magnitude of increasing loading rates (Race et al., 2000; Kemper et al., 2007). Recent detailed investigation of the creep response described the initial displacement of the disc upon loading to be caused by mechanical deformation of the annular, nuclear, endplate and vertebral bone, while longer term displacements were due to fluid flow through the vertebral endplates and the annulus (MacLean et al., 2007; van der Veen et al., 2008; O'Connell et al., 2011b).

The basic mechanical properties of the annulus fibrosus were shown by Brown to depend strongly on location within the disc, with the outermost layers being stiffer and stronger (Brown et al., 1957). In another early study, Galante and colleagues found the annular mechanical properties to depend highly on loading direction (Galante, 1967). Since 1990, additional studies on the annulus fibrosus have confirmed that their mechanical properties depend on location within the disc (Acaroglu et al., 1995; Ebara et al., 1996; Elliott and Setton, 2001), on the direction of loading (Fujita et al., 1997), and that they depend on the level of degeneration in the disc (Acaroglu et al., 1995; Iatridis et al., 1998). The properties of a single lamella from the annulus fibrosus have been determined with lamellae from outer disc regions being 2–3 times stiffer than those from inner regions (Skaggs et al., 1994; Holzapfel et al., 2005).

The nucleus pulposus exhibits fluid-like mechanical properties, as first described by Keyes and Compere (1932). Some interesting research since 1990 demonstrated that this characterization of the nucleus is valid at low loading rates, but that it behaves more solid-like at high loading rates (Iatridis et al., 1996). Further, with aging and degeneration, this group showed how the nucleus pulposus behaves mechanically more like a solid, even at low loading rates (Iatridis et al., 1997).

5.3. *Functional biomechanics*

The classic work of Hirsch and Nachemson showed us that under an axial compressive load the nucleus develops an internal hydrostatic pressure that is contained by the vertebral end-plates and the tension of a bulging annulus fibrosus. With aging and degeneration, the intradiscal pressure decreased and the increasing compressive load was transmitted through the annulus fibrosus. The classic finite element models of Shirazi-Adl were important contributors to this early knowledge (Shirazi-Adl et al., 1984). Since 1990, our understanding of disc loading has been refined by the stress profilometry measurements of Adams and colleagues for both normal and degenerated discs (McNally and Adams, 1992) as outlined in the FSU section. An important study by Krismer and colleagues demonstrated the importance of the annulus fibrosus to torsional stiffness, in addition to the articular facets (Krismer et al., 1996).

The nutritional supply to the intervertebral disc through diffusion across the vertebral endplates was described by Urban and colleagues in the late 1970s through early 1980's (Urban et al., 1977, 1982; Holm et al., 1981). However, it was not until the decade after 2000 and some important clinical and animal observations (Rajasekaran et al., 2004; Lotz and Chin, 2000) that biomechanical studies helped explain the mechanisms for such diffusion, mostly through mathematical modeling (Sélard et al., 2003; Ferguson et al., 2004; Shirazi-Adl et al., 2010; Malandrino et al., 2011). An intriguing aspect of this research is the reproduction of normal diurnal changes in intradiscal pressure (Wilke et al., 1999) using computational and organ culture models (Gantenbein et al., 2006; Chan et al., 2011).

Intervertebral disc damage is a hallmark of aging and degeneration. Other than disc herniation and endplate fracture, which have been discussed previously in the FSU section, other damage includes annular delamination and annular tears. Various patterns of annular tears have been documented in human histopathological specimens (Vernon-Roberts et al., 2007, 2008), but these remain challenging to reproduce in a laboratory environment. One study on human specimens noted that the presence of annular tears decreased the torsional stiffness significantly (Thompson et al., 2000). Annular delamination is presumed to be caused by high annular shear stresses due to increased annular loading as part of the degenerative cascade (Goel et al., 1995; Meakin et al., 2001; Qasim et al., 2014). Further research on these aspects of disc damage is likely important.

5.4. *Intervertebral disc—future steps*

Intervertebral disc research is progressing at a rapid pace as we continue to learn more about its detailed microstructural anatomy and physiology. These insights will surely inform our current understanding of normal disc function and may shed light on injury and damage mechanisms. Future research on disc nutrition is clearly important, particularly in differentiating normal from potentially painful degeneration. Ultimately, the link between biomechanical changes in the disc and clinically relevant changes are important and currently poorly understood (Table 4).

6. Component—spinal ligaments

The spinal ligaments are uniaxial structures that connect adjacent vertebrae along the spinal column. They enable the spine to move within certain limits to avoid damage to the surrounding

Table 5
Spinal ligaments summary. This table presents an overview of our past (up to 1990) and current (1990–2015) knowledge on the key topics related to the spinal ligaments. Possible future research directions are noted in the right-hand column. Short forms used: BMD for bone mineral density.

	Up to 1990	1990–2015	Looking towards the future
Ligament anatomy	i. Basic anatomy of spinal ligaments ii. Little quantitative anatomical data available	i. Further anatomic description of some spinal ligaments ii. Quantitative anatomic data on some spinal ligaments (i.e. cervical) iii. Basic description of mechanoreceptors in spinal ligaments	i. More quantitative data on spinal ligament anatomy, esp. thoracic and lumbar ii. Refine understanding of proprioceptive role of spinal ligaments
Ligament physical properties	i. Non-linear behavior of lumbar spine ligaments described	i. Load-displacement behavior of more ligaments, notably from thoracic and cervical regions ii. Viscoelastic behavior of spinal ligaments iii. Effect of sub-failure injury on ligament physical properties	i. Additional studies needed on basic physical properties? ii. Loading rate sensitivity of various spinal ligaments iii. Further characterization of the effects of sub-failure injuries
Ligament—failure behavior	i. Basic understanding of ligament failure ii. Failure location (i.e. insertion site or mid-substance) believed due to loading rate	i. Failure location found to also be dependent on skeletal maturity	i. Further research needed on failure behavior/sites of spinal ligaments? ii. Role of capsular ligament injury on pain
Ligament—adaptive nature	i. Ligament adaptation to applied loads not well understood	i. Ligament stiffness and strength related to vertebral BMD, suggesting adaptation	i. Basic mechanisms through which the ligaments adapt is important

neurologic structures. A summary of the topics addressed here for the spinal ligaments is shown in Table 5.

6.1. Quantitative anatomy

The basic qualitative anatomy of the spinal ligaments was well known in 1990. In addition to the high collagen percentage, spinal ligaments also contain various degrees of elastin, proteoglycan, and water. The collagen orientation generally runs parallel to the axial direction of the ligament substance, with wider ligaments such as the interspinous ligament having a more variable collagen structure (Aspden et al., 1987; Hukins et al., 1990; Hindle et al., 1990). The presence of nociceptors in spinal ligaments was described as far back as 1949 (Wiberg, 1949; Korkala et al., 1985; Giles and Harvey, 1987).

Since 1990, several studies have described the presence of mechanoreceptors in spinal ligaments (Rhalmi et al., 1993; Yahia and Newman, 1993; Kiter et al., 2010), thereby suggesting an active role beyond as simple elastic stabilizers of the spinal column. Some quantitative anatomy of spinal ligaments has been described for spinal ligaments, consisting of lengths and orientations in the cervical spine (Panjabi et al., 1991c, 1991d). Some similar data exists for the thoracolumbar spine (Jiang et al., 1994), but the incomplete nature of the quantitative data makes it an area for future research.

6.2. Physical properties

The basic non-linear shape of the ligament load–displacement curve was well established in 1990, particularly for lumbar spine ligaments (Tkaczuk, 1968; Nachemson and Evans, 1968; Chazal et al., 1985; Dumas et al., 1987). In the past 25 years, the properties for many more spinal ligaments have been described, including many from the cervical and thoracic regions (Pintar et al., 1992; Jiang et al., 1994; Neumann et al., 1992, 1994a; Yoganandan et al., 1989, 2000; Ivancic et al., 2007; Mattucci et al., 2012; Winkelstein et al., 2000).

Considerable research since 1990 has documented other properties of spinal ligaments than the simple quasi-static behavior referred to above, including the viscoelastic characteristics and some features related to ligament damage. With respect to viscoelasticity, or time-dependent behavior, several studies have documented the effect of loading rate on ligament behavior with higher rates leading to stiffer load–displacement behavior (Lucas et al., 2008; Mattucci et al., 2012). Studies on the load relaxation characteristics of spinal ligaments demonstrated a non-linear viscoelastic feature, whereby the amount of relaxation is dependent upon the amount of initial stretch (Yahia et al., 1991; Lucas et al., 2008; Troyer and Puttlitz, 2012).

Several studies on ligaments from various joints of the body described the load–displacement behavior after a sub-failure stretch as typically involving a longer toe region and a decreased overall stiffness (Panjabi et al., 1996; Pollock et al., 2000; Provenzano et al., 2002). We include these studies here as the observations are likely germane to the post-injury characteristics of the spine.

6.3. Functional biomechanics

Various studies have shown that the physical properties of spinal ligaments change over time, presumably in response to the mechanical loads to which they are subjected. For example, Neumann and colleagues showed that the strength of the anterior longitudinal ligament is correlated with the bone mineral density of the adjacent vertebra (Neumann et al., 1993, 1994b). Tzaczuk (1969) observed ligament strength to decrease with age in the lumbar spine and the same observation has recently been made for the cervical spine (Mattucci et al., 2012). Ligament stiffness and strength were observed by Kotani and Cunningham (1998) to decrease when in parallel with spinal instrumentation, likely a demonstration of disuse atrophy.

6.4. Spinal ligaments—future steps

It is clear that spinal ligaments are more complex than being 'elastic bands', as they are sometimes described. Future research should focus on their adaptive nature and on their proprioceptive characteristics. Further, more detailed quantitative anatomy of spinal ligaments would be a positive contribution to the literature, given the future importance of computational models of the spine.

7. Overview

The last 25 years has seen an explosion in the volume of new research in the field of biomechanics as it relates to the human spine. It is hoped that this review places the novel findings regarding the fundamentals of spine biomechanics into the context of what was known in 1990. While conducting new research is always important, it is critical that we acknowledge and build on the efforts of past investigators. Only in this manner will true, meaningful progress be made.

These are exciting times for researchers in spine biomechanics. We have extensive experimental and computational tools at our disposal to address basic science and applied research questions. Many problems of the spine such as low back pain through spinal trauma undoubtedly will and indeed must include a biomechanical knowledge base in forging new solutions.

Conflict of interest

The author has no conflict of interest related to the content of this manuscript.

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